Department of Veterans Affairs

Journal of Rehabilitation Research and Development

SUMMER 1990

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Rehabilitation Research and Development Service

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Department of Veterans Affairs

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Authors and readers of the *Journal of Rehabilitation Research and Development* are deeply indebted to the *Journal's* referees. The referee process has been fundamental to the quality of the *Journal*. Our policy of anonymity in the review process is intended to help ensure objectivity, but carries with it an unfortunate lack of public recognition. Our referees of national and international stature deserve recognition.

Each review requires hours of intensive analysis and often yields pages of detailed criticism and suggestions. These critiques are passed on (anonymously) to the authors. Without exception, the authors have used the critiques to improve those articles which have been accepted for publication. The result is a journal of high quality, which has expanded its circulation by 500 percent since 1983.

Quality is not an abstraction. Through his or her critiques, each referee has made tangible contributions to the fruitfulness of rehabilitation research and to the well being of veterans and other individuals with disabilities.

To our referees, well done and thanks.

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Effect of velocity and SF/SL ratio on external work and gait movement waveforms

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Abstract—Ten normal adult subjects were tested under a wide range of combinations of stride length and stride frequency. Both longitudinal and vertical accelerations of the body increased with velocity, while stride frequency to stride length ratio (SF/SL) had little effect. The peak-to-peak amplitude of the forward velocity (V) decreased with SF/SL, but at constant SF/SL remained essentially unchanged throughout the speed range from 0.4 to 1.4 m/s. The external work of locomotion related as a paraboloid of revolution to V and SF/SL with minimum work being 0.32 J/kg-m at V = 0.52 m/s and SF/SL = 0.76. The efficiency of transfer between potential and kinetic energies related as a paraboloid of revolution to V and SF/SL with maximum efficiency being 0.58 (58 percent at V = 0.98 m/s and SF/SL = 0.43). Latent work, herein defined as work divided by efficiency, related as a paraboloid of revolution to V and SF/SL being minimum at V = 0.7 m/s and SF/SL = 0.65. At any given velocity, latent work tended to be least in the SF/SL range of 0.6 to 0.7. The latter finding may suggest why people tend to walk at a nearly constant SF/SL ratio over a broad range of velocities.

Key words: *gait analysis, SF/SL ratio, stride frequency, stride length, velocity.*

INTRODUCTION

Fundamental factors of biomechanics come into play which help characterize an individual's gait pattern. Among such factors are the cadence and step length which result in a given forward velocity of walking. Molen (5) and Herman (4) discussed strategy of body movement control when people walk over a broad range of velocities. They observed that an individual adjusts both stride length and stride frequency in almost equal proportions to achieve a desired velocity. It is common knowledge that stride frequency (SF) and stride length (SL) are proportional to the square root of velocity (V); that is, $SF = (V \times a)^{0.5}$ and $SL = (V \div a)^{0.5}$. The proportionality factor a is SF/SL, the stride frequency divided by stride length. A constraint imposed on the locomotor process, such as may result from injury, stroke, or amputation, causes an alteration in movement patterns. Such alterations are reflected in both the time (frequency) and displacement (stride length) domains. Relating gait variables to the product and the quotient of these factors may offer additional means to discriminate subtle differences in means of intervention.

This project addressed the locomotor performance of normal, adult subjects and was conducted as part of an analysis of the effect on gait of a variety of prosthesis suspension systems for below-knee amputees. The purpose of this project was to investigate the effect of velocity and SF/SL ratio on selected gait movement waveforms and on external work and to provide comparative data in the analysis of amputee gait.

METHODOLOGY

Instrumentation

Conventional body contact instrumentation was used to collect locomotion data. A total of nine channels were

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This work was supported by a grant entitled "Analysis of Below-Knee Suspension Systems: Effect on Gait," from the Department of Veterans Affairs Rehabilitation Research and Development Service and was conducted at the Department of Veterans Affairs Medical Center, 3350 La Jolla Village Drive, San Diego, CA 92161.

monitored; these included bilateral heel and toe switches, bilateral knee electrogoniometers, a biaxial accelerometer, and a tachometer. In addition, the subjects wore a lightweight backpack containing plugs for connections to a trailing cable and a snug-fitting hip-waistband for the accelerometer mount and for attaching the tachometer cable. Signals were conveyed to the signal conditioning console via an overhead trailing cable.

The foot switches, about 2 mm thick and consisting of a piece of conductive silicone rubber sandwiched between two brass shim stock electrodes, were taped to the heel and toe area of the footwear. The electrogoniometers were mounted anteriorly on the limbs to register knee flexion and extension. Elastic straps secured mounting pads to the shin and thigh areas, while a pair of parallel linkage arrangements transferred relative angular limb segment movements to a potentiometer.

The tachometer was a low inertia dc generator mounted on a stationary base at one end of the walkway. A stranded steel cable, 0.014-inch in diameter, passed from a supply reel around a pulley mounted on the generator shaft and attached to the hip-waistband worn by the subject. A weight and pulley system maintained a constant tension of about 1 newton on the cable.

The accelerometer was an Entran triaxial device used in a biaxial mode. The accelerometer was mounted on the hip-waistband and housed in a 2-axis gimbal arrangement to allow stabilization of the "X" axis to the laboratory reference in the direction of body progression. Stabilization was done by coupling to the tachometer cable through a mechanical filter. The filter consisted of a thin slice of soft plastic foam about 1-inch square. A slit to the center of the piece of foam allowed convenient attachment to and detachment from the tachometer cable. This scheme attenuated mechanical noise arising from the cable.

The walkway was a tiled floor surface approximately 8 meters long, of which a distance of 6 meters was used as the test zone. An overhead track supported a flat ribbon cable of twisted pairs of leads to carry the signals to the console. The drag of the trailing cable was about 1 newton. The signal conditioner amplified and filtered the signals. A microcomputer system sampled each channel at 60 times per second and performed the analog-to-digital conversion with 8-bit resolution. Data were recorded on diskettes for processing.

Test procedure

Ten subjects were included in the study: eight males and two females, ranging in age from 23 to 63, in height from 153 to 196 cm, and in weight from 52 to 91 kg. The test protocol provided for about 15 test walks for each subject in the approximate velocity range from 0.3 to 1.5 meters per second and in the approximate SF/SL ratio range from 0.3 to 1.1.

Table 1.
Characteristics of test subjects.

Subj	Ht *cm*	Wt *kg*	Age *yrs*	Sex	Vo *m/s*	SFo *l/s*	SF/SLo	n
1	175	84	62	M	1.12	0.85	0.65	18
2	153	57	38	M	1.25	0.87	0.61	15
3	163	52	23	F	1.00	0.85	0.72	17
4	196	82	23	M	1.33	0.82	0.51	12
5	188	91	63	M	1.03	0.81	0.63	17
6	179	80	54	M	1.27	0.87	0.59	15
7	183	84	41	M	1.07	0.89	0.75	15
8	191	82	23	M	1.01	0.78	0.61	18
9	173	68	49	F	1.18	0.95	0.77	14
10	175	68	60	M	1.09	0.82	0.62	14

Vo = Velocity, first test
SFo = Stride frequency, first test
SF/SLo = Ratio stride frequency to stride length, first test
n = number of tests contributed

After instrumentation for the first test, each subject was instructed to walk at a free choice of cadence and speed. This became the first test condition and was the point of departure (see **Table 1**) for the balance of the testing for each individual. Four target velocities in the vicinities of 125, 100, and 60 percent of the first test speed, and a slow speed in the order of 0.3 to 0.4 m/s constituted the speed range. These speeds were reached with practice walks which were monitored on a storage oscilloscope. At each target velocity (perceived subjectively by the subject) individuals were tested walking at their comfortable choice of stride frequency and stride length. Subsequently, by taking longer steps at lower cadences or shorter steps at higher cadences at those velocities, respectively, target SF/SL ratios were achieved.

To get the subjects to walk at the very slow speeds, they were asked to imagine themselves accompanying a recent stroke patient just learning to ambulate with a walker. Once the low speed was achieved, subjects were instructed to walk at a comfortable choice of cadence and step length for a test bout. For the low and high SF/SL conditions, the subjects were instructed to walk at that same slow speed (perceived subjectively), and to take long steps at low

cadence and short steps at high cadence to provide the low and high SF/SL ratios, respectively.

A storage oscilloscope was used to display the measured waveforms for visual inspection to ascertain that the desired test condition had been achieved and that the data were suitable for recording. If data were deemed unsuitable, the data were not recorded and the test was repeated.

Data treatment

The beginning of the gait cycle was chosen to occur when the left knee reached peak extension at the conclusion of the swing phase. This event was confirmed by the subsequent closure of the heel switch. The average velocity of each gait cycle was examined to cull out those cycles clearly accelerative after initiation of gait, and those clearly decelerative prior to cessation of gait.

Measured were velocity and cycle time. The reciprocal of the cycle time was used to derive the stride frequency. The average velocity, determined for the gait cycle, was multiplied by the gait cycle time to derive the stride length. The SF/SL ratio was derived by dividing the stride frequency by the stride length.

Waveform Analysis. The data for the tachometer and accelerometer waveforms were quantitated using the Fourier series technique. Fourier coefficients for the first 12 harmonics were calculated using procedures as described in textbooks and handbooks. Basically the gait cycle, of time duration T, was divided into 24 equal parts at time intervals *t*. The cosine and sine coefficients, A_n and B_n, were calculated by the following:

$$A_n = 2/T \int_O^T F(t)\cos(2\pi nt/T)\,dt$$

$$B_n = 2/T \int_O^T f(t)\sin(2\pi nt/T)\,dt \qquad [1]$$

where:

$F(t)$ = magnitude of wave form at time t
n = value of the harmonic being solved.

Harmonic ratios (HR) were calculated for the waveforms in the manner described by Robinson (6). The HR is the sum of the absolute values of the even harmonics divided by the sum of the absolute values of the odd harmonics:

$$HR = \frac{\Sigma(A_e + B_e)}{\Sigma(A_o + B_o)} \qquad [2]$$

where:

A_e and B_e = absolute values of the even harmonics
A_o and B_o = absolute values of the odd harmonics.

The Fourier coefficients and HR derived for each gait cycle in a given test condition were averaged to represent the best estimates of performance. Similarly, average velocities and SF/SL ratios were derived to represent each test condition. Multiple regression was used to relate the averaged Fourier coefficients to V and SF/SL.

External Work. The calculation of external work of locomotion was based on the pioneering effort by Cavagna (1). The procedure consisted of incrementally advancing through the gait cycle by sampling time intervals to calculate changes in g's acceleration into meters per second squared. Because of small drift accumulated during each gait cycle, slope corrections were applied to the acceleration values to render the gait cycle to a net zero gain. Velocity changes during the gait cycle for both the forward and vertical directions were derived by multiplying acceleration *a* by the sampling time interval t_i: $\Delta V = at_i$. The average forward velocity, Va, derived from the tachometer, was applied to the forward velocity increments to provide the laboratory inertial reference for calculation of kinetic energies. Kinetic energy changes were derived by the following:

$$\Delta KE = 1/2\,(V_2^2 - V_1^2) \qquad [3]$$

Changes in the vertical displacement were derived by multiplying the vertical velocity V_v by the sampling time interval t_i: $\Delta h = V_v t_i$. Potential energy changes were derived by the following:

$$\Delta PE = g(\Delta h) \qquad [4]$$

where g is the acceleration of gravity.

For each gait cycle, the values shown in brackets were accumulated only if they were positive:

$$\begin{aligned} \Delta KE &= [\Delta KE_l + \Delta KE_v] \\ \Delta PE &= [\Delta PE] \\ \Delta W &= [\Delta KE_l + \Delta KE_v + \Delta PE] \end{aligned} \qquad [5]$$

where:

ΔKE_l = change in kinetic energy in the longitudinal direction
ΔKE_v = change in kinetic energy in vertical direction
ΔPE = change in potential energy [6]

The efficiency of transfer between potential and kinetic energies is the following:

$$E = [1 - \frac{\Sigma W}{\Sigma KE + \Sigma PE}] \qquad [7]$$

Work, ΣW, derived from accelerometry is work per unit mass for the gait cycle. To normalize into units of work

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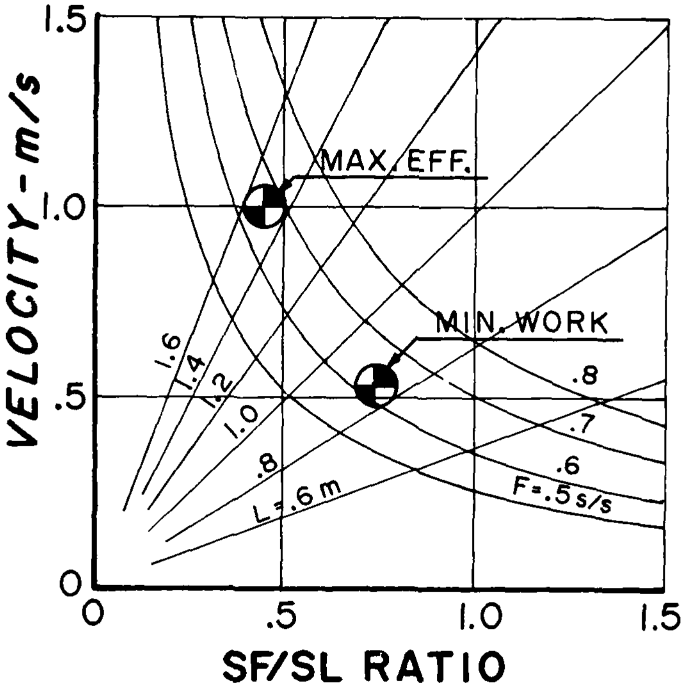

Figure 1.
Minimum work and maximum efficiency on a plot of velocity versus SF/SL ratio.

per unit mass per unit distance walked, the resulting summed work ΣW was divided by the product of average velocity Va and cycle time T for the gait cycle:

$$\text{Work} = \Sigma W/(Va \times T) \qquad [8]$$

Latent work was calculated using the derived values for work and efficiency on the rationale that external work equals the product of latent work and efficiency. Hence, latent work as defined in this report, was derived by dividing external work by efficiency.

Regression analysis methods were used to empirically relate work and efficiency to velocity V and ratio of stride frequency to stride length SF/SL. Both work and efficiency were found by approximation to be related parabolically to V and SF/SL. Since the minimum work and maximum efficiency occurred within the V versus SF/SL field, parabolic curve fitting was done by adjusting assumed vertex values of V and SF/SL for best fit. Optimization was done by using the highest correlation coefficient as the criterion for best fit. After the best estimated coefficients were derived, test points from the data pool were tested against the derived formula for the amount of scatter and expressed as a standard deviation.

RESULTS

A total of 155 tests points were obtained among the ten subjects tested. This database was treated as a pool to yield trends as affected by the various combinations of velocity V and SF/SL ratio.

Work and efficiency

Work and efficiency related to V and SF/SL as paraboloids of revolution. The empirical expressions for work and efficiency were the following:

$$W = 0.32 + 0.38(0.52 - V)^2 + 0.42(0.76 - SF/SL)^2$$
Std.Dev. = 0.085 J/kg-m

$$E = 0.58 - 0.36(0.98 - V)^2 - 0.30(0.43 - SF/SL)^2$$
Std.Dev. = 0.104 (10.4 percent) [9]

The minimum work was 0.32 J/kg-m at V = 0.52 m/s and SF/SL = 0.76. The maximum efficiency was 0.58 (58 percent) at V = 0.92 m/s and SF/SL = 0.43 (see **Figure 1**).

The minimum latent work was 0.63 J/kg-m at V = 0.7 m/s and SF/SL = 0.65. The latent work distributed about the minimum as a paraboloid of revolution according to the following empirical expression:

$$Wl = 0.63 + 1.2(0.7 - V)^2 + 1.2(0.65 - SF/SL)^2 \qquad [10]$$

At any given velocity, latent work tended to be least in the SF/SL range of 0.6 to 0.7 (see **Figure 2**).

Potential and kinetic energy waveforms, available from the work calculations, were examined. These showed effects of both V and SF/SL. At velocities and SF/SL ratios comfortable to the subjects, the waveforms were relatively smooth and approximately sinusoidal with the kinetic and potential energy curves phased about 180 degrees apart. During double support period, the kinetic energy was maximum while the potential energy was minimum. At midway between double support periods, the kinetic energy was minimum and the potential energy was maximum. The peak-to-peak amplitudes increased with velocity and decreased with SF/SL. At low velocities, and at both low and high SF/SL ratios, the waveforms were distorted and showed evidence of phase shift between the kinetic and potential energy curves. There was no apparent consistency in whether the kinetic energy led or lagged the potential energy. Causes for the distortions and phase shifts could not be determined.

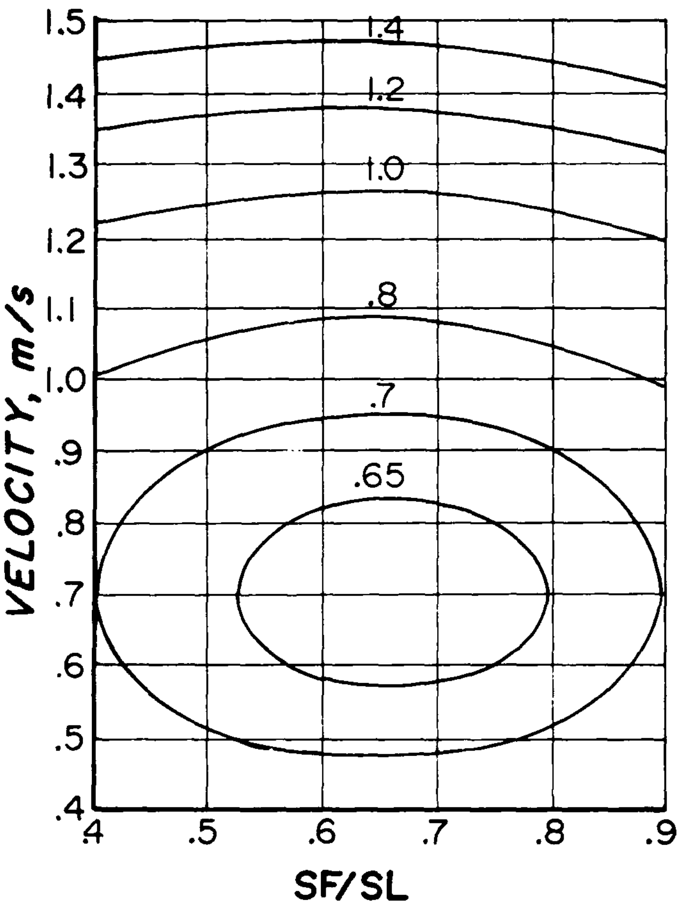

Figure 2.
Latent work, in J/kg-m, on a plot of velocity versus SF/SL.

Waveforms

The coefficients related linearly to velocity V and SF/SL ratio in the following general form:

$$FC = B1(V) + B2(SF/SL) + C \qquad [11]$$

This allowed synthesis of representative waveforms at any desired combination of velocity V and ratio of stride frequency to stride length SF/SL. For illustrative purposes, curves were synthesized at the following conditions:

a. V = 0.6, 1.0, and 1.4 m/s at SF/SL = 0.6
b. SF/SL = 0.3, 0.6, and 0.9 at V = 1.0 [12]

Briefly, the effects of velocity V and the ratio of stride frequency to stride length SF/SL are summarized as follows:

1. Accelerometer

Velocity had the predominant effect on both the longitudinal and vertical acceleration waveforms while the SF/SL ratio had only minor effects.

The peak-to-peak amplitude of the vertical acceleration increased from ±0.11 g at 0.6 m/s to ±0.25 g at 1.4 m/s. The SF/SL had essentially no effect on the shape of the vertical acceleration waveform.

The peak-to-peak amplitude of the longitudinal acceleration increased with velocity from ±0.12 g at 0.6 m/s to ±0.25 g at 1.4 m/s. The longitudinal acceleration waveform also changed shape with SF/SL. At SF/SL = 0.3, the waveform resembled a sawtooth. At SF/SL = 0.9, the decelerative components were of short duration, and the accelerative components were almost constant (see **Figure 3**).

2. Tachometer

The SF/SL ratio had the predominant effect on the tachometer waveform. The peak-to-peak amplitude decreased with increasing SF/SL from ±0.21 m/s at SF/SL = 0.3 to ±0.11 m/s at SF/SF = 0.9. Velocity had a lesser effect on the amplitude. The peak-to-peak amplitude increased with average forward velocity from ±0.13 m/s at 0.6 m/s to ±0.19 m/s at 1.4 m/s (see **Figure 3**).

Harmonic ratios

The means and standard deviations of the HR, derived from the pooled data of the 155 tests, for the longitudinal acceleration (LACC), the vertical acceleration (VACC), and the tachometer (TACH) were the following:

	Mean	Std. Dev.	
HR LACC	4.42	1.90	
HR VACC	3.10	1.24	
HR TACH	4.41	1.87	[13]

Harmonic ratios were influenced by both velocity and ratio of stride frequency to stride length. The HR increased with velocity for the two accelerations and decreased for the tachometer. The HR decreased with SF/SL ratio for the accelerations and tachometer waveforms. The magnitude of the effects are summarized as follows:

HR LACC:
At SF/SL = 0.6, HR HACC increased from 4.44 at V = 0.6 m/s to 4.60 at V = 1.4 m/s.
At V = 1.0 m/s, HR HACC decreased from 5.04 at SF/SL = 0.3 to 4.01 at SF/SL = 0.9.

HR VACC:
At SF/SL = 0.6, HR VACC increased from 2.84 at V = 0.6 m/s to 3.57 at V = 1.4 m/s.
At V = 1.0 m/s, HR VACC decreased from 3.33 at SF/SL = 0.3 to 3.08 at SF/SL = 0.9.

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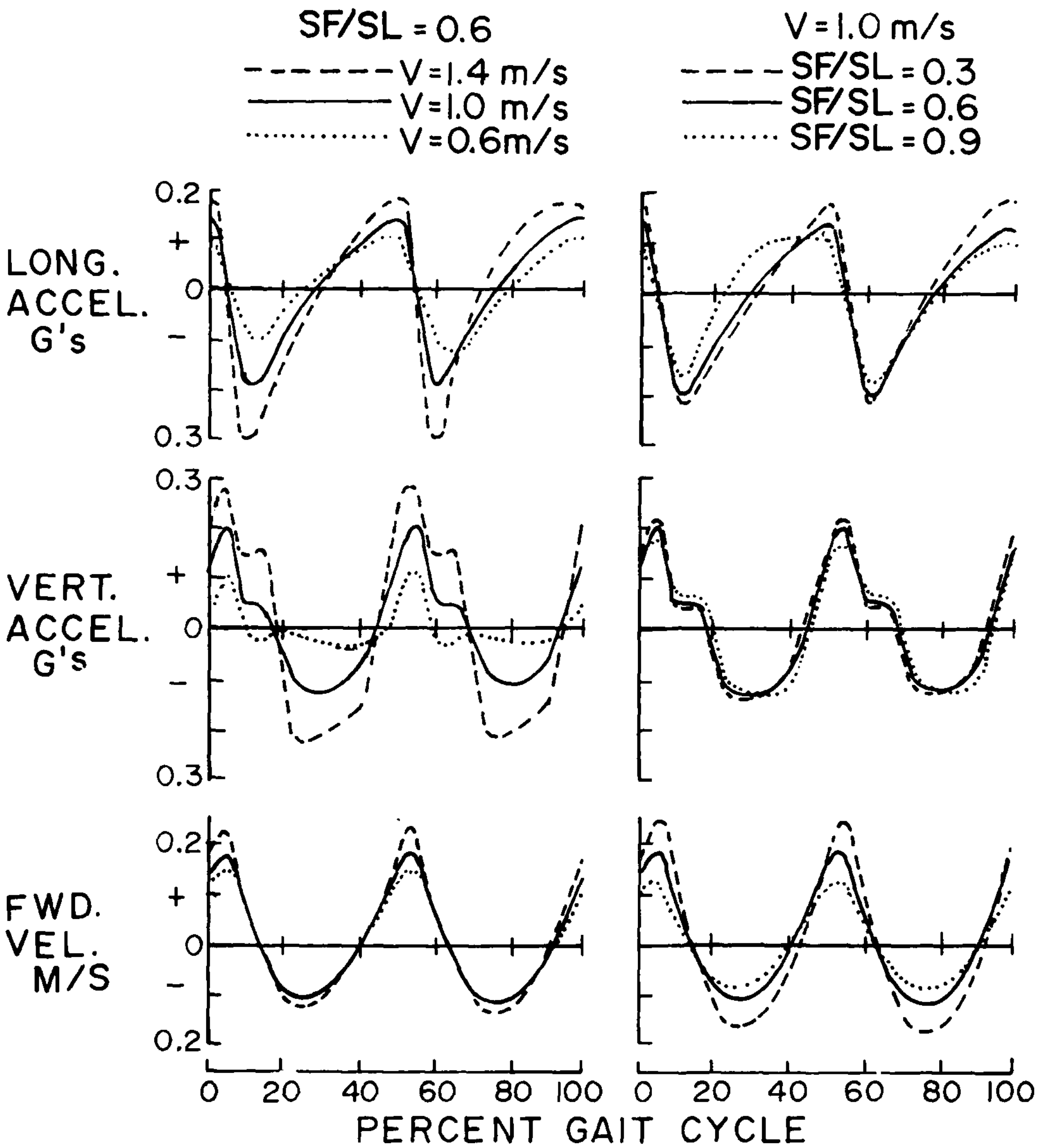

Figure 3.
Effects of velocity and SF/SL ratio on waveforms.

HR TACH:

At SF/SL = 0.6, HR TACH decreased from 5.00 at V = 0.6 m/s to 3.70 at V = 1.4 m/s.

At V = 1.0 m/s, HR TACH decreased from 5.17 at SF/SL = 0.3 to 3.54 at SF/SL = 0.9.

[14]

DISCUSSION

The minimum work did not coincide with maximum efficiency in values of velocity and SF/SL ratio. Both minimum work and maximum efficiency occurred at about the same stride frequency, but at different stride lengths. It should be no surprise that minimum work and maximum efficiency occur at the same frequency, since the body consists of an assembly of joined segments which can be viewed as time-regulating pendulums. One can speculate that the body resembles a low Q system with minimum impedance at resonant frequency, supporting the notion that efficiency reflects the degree of coordination of movements of various parts of the body. Craik (2) investigated relative bilateral flexion-extension movements of the hips and shoulders, and found that at higher velocities the movements were diagonal (reciprocal), while at lower velocities they were homologous. The transition zone occurred at a stride frequency of about 0.75 Hz. Moreover, Craik found

phase differences to exist between the hip and shoulder movements, depending on whether the subject walked above or below the transition zone frequency. The shapes of potential and kinetic energy curves suggested smooth functioning at comfortable cadences and step lengths. When the subjects walked at conditions alien to their comfortable choices of stride frequency and stride length, the curves showed distortions and evidence of phase shifts suggesting less than optimum coordination among segments of the body. Since movements of the arms and torso were not measured, causes for the distortions and phase shifts could not be determined.

The manner in which latent work distributed with respect to velocity and ratio of stride frequency to stride length may suggest why people walk with a relatively constant SF/SL ratio over a wide range of speeds. One could argue that the body adjusts both stride frequency and stride length in equal proportions to economize in energy.

The tachometer waveform, representing the forward velocity of the body, showed sensitivity to SF/SL because of relative step length. At high SF/SL, step lengths are short, resulting in a small vertical excursion of the body center of gravity. Conversely, at low SF/SL, the step lengths are long, and consequently, the vertical excursion of the body center of gravity is large. The important feature in the tachometer waveform was that the peak-to-peak amplitude was relatively unchanged as a function of velocity. This suggested a significance as noted by Herman (4) in that the body may regulate the cyclic behavior by a nearly constant variation in *momentum*. This leads to consideration regarding stability in locomotion. The slow walks were the most difficult for the entire group. At higher average forward velocities, the minimum momentum is large enough that a small perturbing impulse does not threaten the stability of the moving body. Conversely, at low velocities, the minimum momentum is so small that a similar small perturbing impulse could be a threat.

Variation in amplitude of the longitudinal acceleration as a function of velocity is a reflection of the slopes of the tachometer waveform when viewed as a function of time. While the peak-to-peak amplitude of the forward velocity remained essentially constant throughout the velocity range, the slopes varied inversely as the duration of the gait cycle. Variations in the vertical acceleration reflect the rate of change in the vertical velocity of the body center of gravity which is conditioned by both step length and step frequency.

The overall mean values of harmonic ratios for the group of normals exceeded 3.0 which is in agreement with the findings of Robinson (6). Gage (3) noted that in the harmonic analysis of vertical acceleration, changes in relative magnitudes of the odd harmonics were attributable to bilateral asymmetries as follows: uneven toe-out depressed the third harmonic, uneven arm swing elevated the fifth harmonic, and lateral trunk bending elevated the seventh harmonic. No subjects in these tests demonstrated substantive increases in bilateral asymmetry as a result of test conditions.

The choice of 12 harmonics to quantitate waveforms was based on practical assessments of expected utility. Twelve harmonics were more than adequate to characterize the tachometer waveform. If one were concerned about sharp spike-like features in the accelerometer waveforms, 12 harmonics would be marginal. In this study, the issues centered on relating work and general waveform features to velocity and SF/SL ratio. Considering the sampling rate and possibilities of motion artifact stemming from instrument mounting, analyses using 12 harmonics were thought to be adequate.

CONCLUSIONS

1. External work of locomotion related to velocity V and ratio of stride frequency to stride length SF/SL as a paraboloid of revolution where minimum work was 0.32 J/kg-m at SF/SL = 0.76 and V = 0.52 m/s. Efficiency of transformation between potential and kinetic energy also related to V and SF/SL as a paraboloid of revolution where maximum efficiency was 0.58 (58 percent) at SF/SL = 0.43 and V = 0.98 m/s. Latent work, defined herein as external work divided by efficiency, related to V and SF/SL as a paraboloid of revolution being its least of 0.63 J/kg-m at SF/SL = 0.65 and V = 0.7 m/s. At any given velocity, latent work tended to be least in the SF/SL range of 0.6 to 0.7.

2. Calculation of Fourier coefficients permitted synthesis of representative waveforms of the vertical and longitudinal accelerations, and tachometer. The following were observed: (a) The peak-to-peak amplitude of the vertical acceleration increased from ±0.11 g at 0.6 m/s to ±0.25 g at 1.4 m/s. The SF/SL ratio had essentially no effect; (b) The peak-to-peak amplitude of the longitudinal acceleration increased with velocity from ±0.12 g at 0.6 m/s to ±0.25 g at 1.4 m/s. At SF/SL = 0.3 the waveform resembled a sawtooth, while at SF/SL = 0.9 the decelerative components were of short duration and the accelerative components were almost constant; and, (c) The forward velocity (tachometer) waveform decreased in its peak-to-peak amplitude with increasing SF/SL from ±0.21 m/s at SF/SL = 0.3 to ± 0.11 m/s at SF/SL = 0.9. The peak-

to-peak amplitude increased only slightly with velocity from ± 0.13 m/s at V = 0.6 m/s to ± 0.19 m/s at V = 1.4 m/s.

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A Technical Note

Load-bearing characteristics of polyethylene foam: An examination of structural and compression properties

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Abstract—Cellular foams used in shoe insoles come in a variety of structural configurations differing with respect to cell geometry. The compression of a cellular foam depends on cell reticulation and size. Examination of the compression behavior of closed cell polyethylene foams revealed distinct time-and nontime-related properties that occur during static and cyclic loading. Physical parameters were developed and determined to exactly specify pressure profiles that occur at the plantar interface. Evaluation of an interface material can be made based on peak pressures which are dependent on depth of compression, foam thickness, and physical properties of the foam. Sustained loading damages the mechanical integrity of the cellular structure such that thickness does not completely recover. Therefore, issuing several pairs of thick insoles for daily rotation is recommended for a particular foam.

Key words: *biomechanics, cellular polyethylene foams, interface materials, shoe insoles.*

INTRODUCTION

The use of cellular foams in the orthotics and prosthetics industries is widespread and ranges from applications as shoe insole material to prosthetic limb inserts. It is our impression that orthotic and prosthetic practitioners select interface materials, including cellular foams, in an arbitrary fashion based on availability and personal knowledge. Contributing to the arbitrariness of material selection is a lack of published information on detailed mechanical properties of cellular foams. In order to provide a basis for a more complete understanding of cellular foam properties, our investigative efforts have been directed toward an examination of the fundamental properties of cellular foams. Specifically, we have studied the structural and compression properties of cellular foams. An application of these findings is presented in an examination of the cushioning characteristics of some polyethylene foams. This discussion is relevant to orthotists and prosthetists because it addresses an overview of the properties of cellular foams, the knowledge of which may be useful in the determination of the function of a particular foam material in load-bearing applications. It should be noted at the outset that we are not a testing laboratory and that the extent of material testing was limited to support of a grant concerned with the prevention and treatment of diabetic plantar ulcers.

STRUCTURE OF FOAMS

Cellular polyethylene foams such as Aliplast, Plastazote, and Pelite are best described as a mass of bubbles composed of a plastic and a gas phase. The polymer is distributed in the walls of the bubbles and the lines where the bubbles intersect (1). The bubbles are referred to as cells, the lines of intersection are called ribs or strands, and the walls are called windows. Cell windows which consist of polymer material are called nonreticulated cell walls; windows with no structural wall are called reticulated cell

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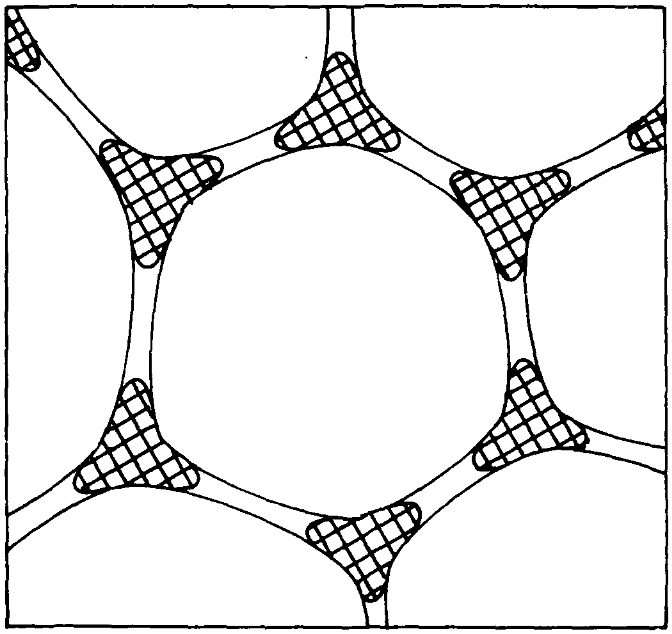

Figure 1a.
A stylized diagram of an open cell structure illustrating cell strands and walls.

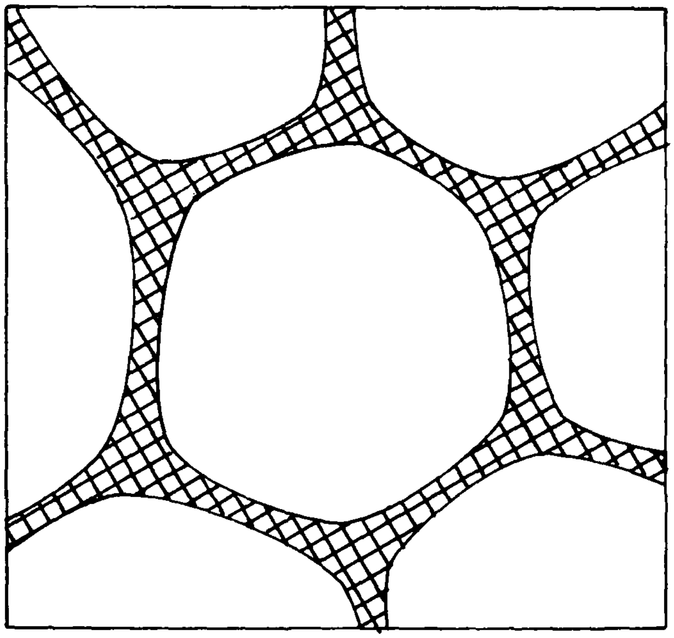

Figure 1b.
A stylized diagram of a closed cell structure illustrating cell strands and walls. A closed cell structure prevents gas movement.

walls. Depending on the configuration of this two-phase gas/solid system and on the synthetic material used, cellular plastics exhibit a wide range of mechanical properties. Hence, for a particular polymer, these properties are controlled by the structural features of the cellular material. In general, two major descriptions are offered to characterize structural features of cellular materials. An open cell material is one which has open windows leaving many cells interconnected in such a manner that gas may pass from one cell to another. Alternatively, closed celled materials are made up of discrete cells through which gases do not pass freely. Stylized diagrams of open and closed cell structures are presented in **Figure 1a** and **Figure 1b**. Note that for the open cell structure, the bulk of the polymer is in the strands. In the closed cell structure, the polymer material is also distributed in the walls, thus preventing gas movement. The open and closed celled classifications of cell structure are important because a knowledge of cell geometry will be desirable in an analysis of physical properties.

PHYSICAL PROPERTIES

One major dependence of mechanical properties on structural variables is demonstrated by compressive loading data. A physical test of the mechanical behavior of a material can be done by continuously measuring the force required to develop a degree of compression. This information is useful because it aids in an evaluation of a foam's response under load-bearing conditions. Results of compression data for polyethylene foam obtained by Skochdopole (2) are illustrated in **Figure 2** which is a plot of compressive load versus percent compression for polyethylene foams of increasing open cell character. The method of compression testing described above was used to obtain compressive load data; load indicating force per unit area. These data show that compressive load of polyethylene foam increases as the fraction of open cells decreases. Consequently, it can be concluded that cell geometry is important in determining the compression characteristics of cellular foams. That is, an interpretation of the Skochdopole results suggests that the distribution of foam material in either the cell ribs or the cell walls influences the compression behavior of the cellular foam. Since this is the case, a consideration of cell geometry is required in the proper use of these materials in load-bearing applications (i.e., as in foot-orthotic devices.)

In order to clarify this result, a model of cellular foam consisting of two reactive elements in parallel was proposed by Skochdopole. In this model, the reactive elements are the cell walls and the trapped gas, as shown in **Figure 3**. From the curve produced for compression of the cell walls, it is observed that at low compressions the resistance of the cell wall dominates. This region then levels off when

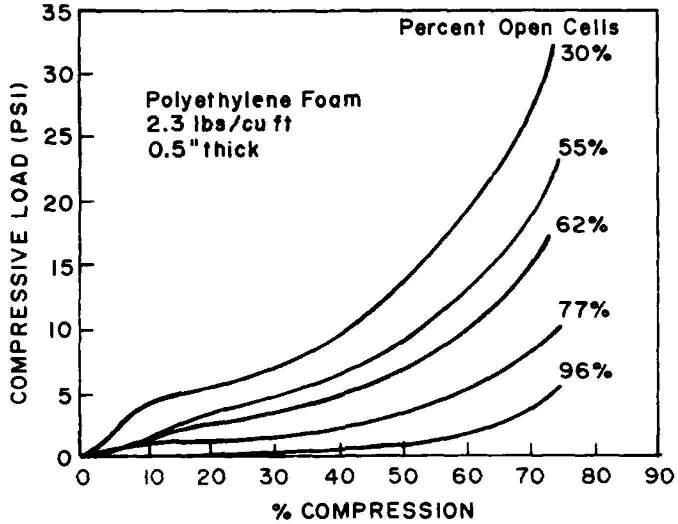

Figure 2.
Compressive load versus compression for different fraction open cells.

the cell walls reach a critical buckling stress and this behavior continues until the matrix is collapsed. At this point, additional loading does not further compress the material. Alternatively, compression of the closed cell element shows that there is a rise in pressure of the trapped gas. Based on this cellular foam model, compression of a foam sample must represent the resultant compressive resistance produced by both elements. In **Figure 2**, the high compressive load upon initial compression for a foam with few open cells may be interpreted as a domination of the compression force by the cell walls. That is, when there is a small fraction of open cells, the compression force is distributed over a larger number of cell walls and ribs, thereby increasing the compressive resistance. At larger degrees of compression, the data presented in **Figure 2** indicates that compressive load increases as the fraction of open cells decreases. The cell geometry model predicts this behavior because gas trapped in closed cells begins to offer increased amounts of compressive resistance as a cellular foam is compressed. This implies that foams of increased open cell character must provide less resistance to escape of gases, which explains the reduction in compression resistance as open cell character increases.

Having determined the effect of reticulation on compression behavior, we can now address the influence of cell size. Knowledge of the influence of cell size on foam compression behavior is important because it is often necessary to compare the loading characteristics of two materials of similar degrees of reticulation. An example would be compression of two closed cell foam samples of similar density which differ slightly with respect to cell size. The larger cell foam illustrated in **Figure 4a** has fewer ribs and walls that undergo compression compared to the small cell foam shown in **Figure 4b.** Hence, if the compression of foam is interpreted as a buckling of cell walls and the intercellular movement of gases, it would appear that the effect of cell size would be significant. The dependence of compression load upon cell size for polyethylene foam is shown in **Figure 5**, reproduced from Skochdopole. Compressive load is plotted versus percent compression for two polyethylene foam samples of similar density which differ considerably with respect to cell size. There is a noted difference in compression behavior at larger compressions. Viewing the ribs of the cellular structure as columns in compression, the columns are shorter in the small cell structure than in the larger cell structure. One could argue that a tall column would buckle sooner under a given load than a short column. This, coupled with higher resistance to intercellular gas flow in a small cell structure, explains the differences between curves A and B in **Figure 5** at larger compressions.

In summary, based on evidence provided by compression data for polyethylene foam, it can be concluded that the influence of cell geometry on the mechanical properties of cellular foams is significant. Specifically, increased compression strength is acquired as the cell diameter decreases. In addition, decreasing the fraction of open cells increases the required force for a given degree of compression. An examination of compression data therefore reveals a relation between structural variables and mechanical

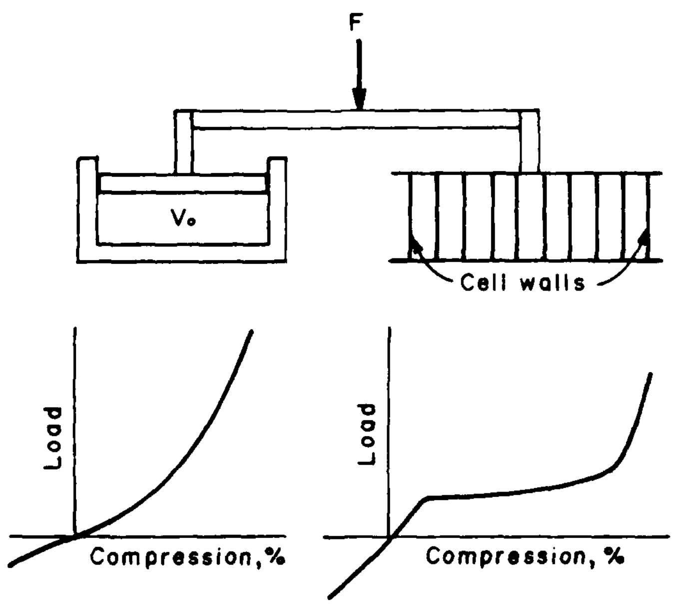

Figure 3.
Mechanical model for polyethylene foam.

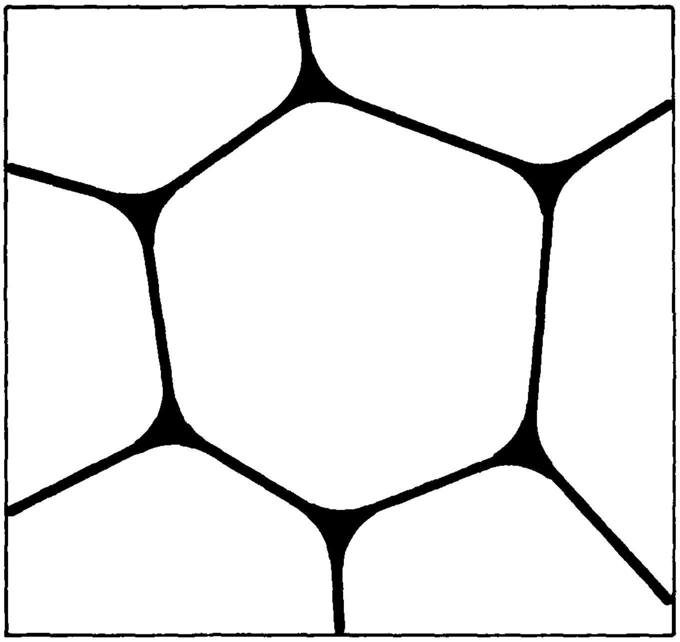

Figure 4a.
A closed cell foam sample of large cells, differing slightly with respect to cell size, but of similar density to the cells shown in **Figure 4b**. Cell size determines the relative number of ribs and walls.

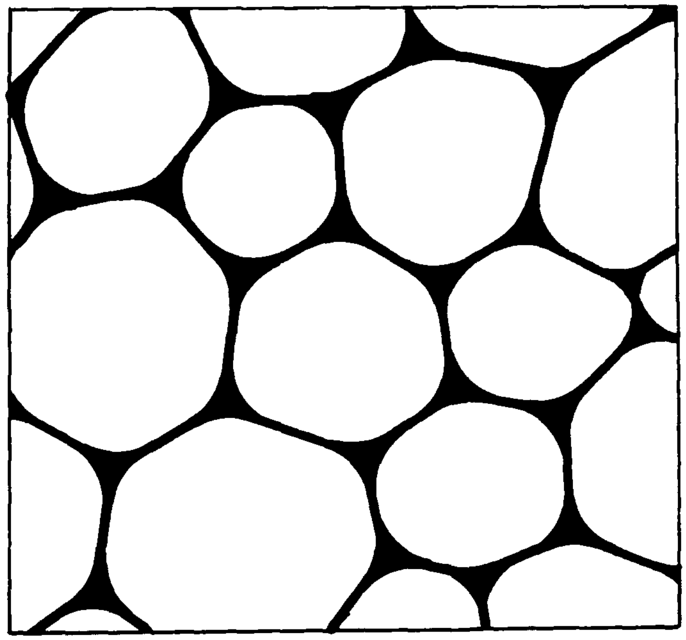

Figure 4b.
A closed cell foam sample of small cells, differing slightly with respect to cell size, but of similar density to the cells shown in **Figure 4a**. Cell size determines the relative number of ribs and walls.

properties for cellular foams: mechanical properties are dependent on the proportion of open cells and on cell size.

With this background information, we can now consider the use of foams under load-bearing conditions with special consideration given to closed cell foams. Closed cell polyethylene foam materials exhibit both time-related and nontime-related properties under load-bearing conditions. The nontime-related properties happen under rapid cyclic loading conditions as may occur during a gait cycle. The time-related properties happen when a load is sustained either as a static load or an extended period of cyclic loading. Under short-duration cyclic loading conditions, the thickness of the material recovers instantly and totally on removal of the load. Under static or prolonged cyclic loading, the material does not return instantly to original thickness on removal of the load. The reason for the latter phenomenon is that, while the structure may be closed cell, the cell walls are not totally impermeable to the flow of gases. Under sustained load, gases are squeezed out; when the load is removed, gases are drawn back into the cells. The recovery is a result of potential energy stored in the cellular matrix.

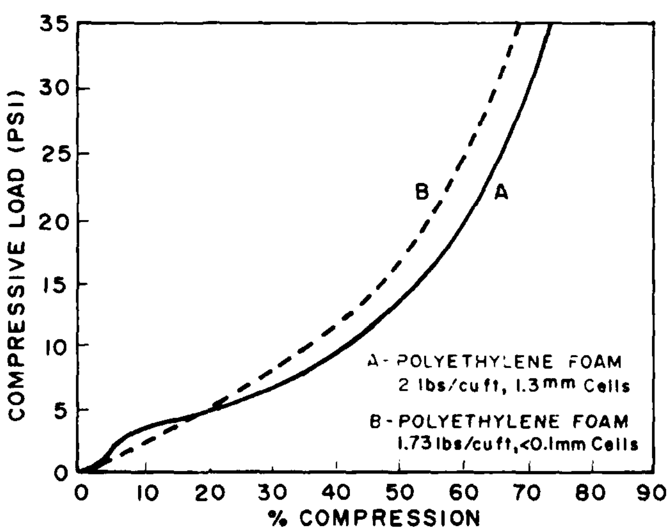

Figure 5.
Compressive load versus compression for different cell sizes.

In particular, nontime-related characteristics resemble a nonlinear spring in load-thickness relations. **Figure 6**, which is a typical thickness versus load relation, illustrates this point. This nontime-related characteristic was determined with a manually operated device which plotted the relation between thickness and the applied force (i.e., pressure). The sample was placed between the jaws of the device, force was applied by a manually operated lever, and an attached pen traced the resulting curve. During loading, energy is stored in both the compression of the gases in the cells and in elastic deformation of the plastic.

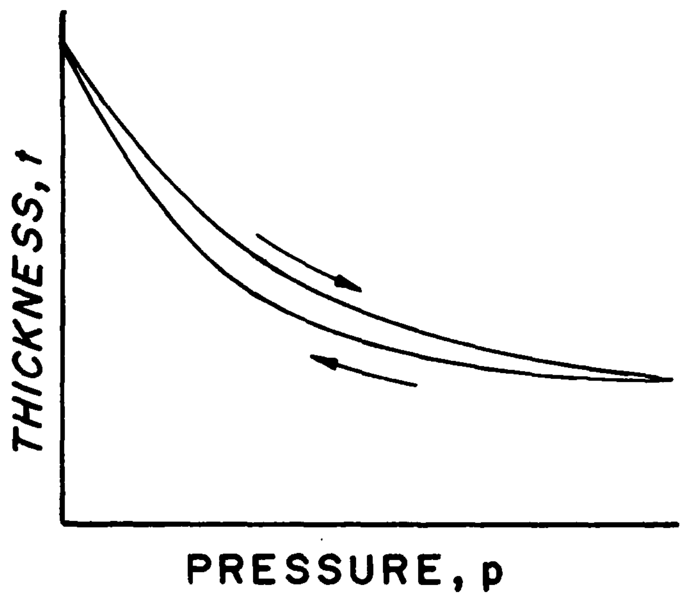

Figure 6.
Typical relation of thickness as a function of applied pressure.

When the load is released, the stored energy restores the thickness of the material to its original value. A small amount of hysteresis may be evident as represented by the loading and unloading paths shown in **Figure 6.**

The relation between the thickness and load may be characterized adequately for discussion purposes by the following approximation:

$$t = (t_0 - t_c)e^{-ap} + t_c \qquad [1]$$

where t is the thickness at any load p, t_0 is the unloaded thickness, t_c is the thickness at large loads where the curve flattens out and approaches a limit, and a represents the compliance factor. The compliance factor may vary from one material to another as well as the relation of t_c to t_0. If the relation of t_c to t_0 is given as $r = t_c/t_0$, then the term r represents the compressibility factor. Substituting r into the above expression results in the following:

$$t = t_0(1 - r)e^{-ap} + rt_0 \qquad [2]$$

The form of this expression shows that a and r are independent parameters.

The effects of different compliance and compressibility factors on the thickness-load relations are illustrated in **Figures 7a** and **7b.** The compliance factor characterizes the rapidity at which the thickness changes with load to reach the limit t_c as shown in **Figure 7a.** The compressibility factor, on the other hand, characterizes the fraction of the unloaded thickness at which the t_c reaches a limit as shown in **Figure 7b.** Values of compliance factor and compressibility factor are listed in **Table 1** for selected commercially available polyethylene cellular foams. The work required in a compression from 0 to 40 psi (276 kPa) indicating the potential energy required to overcome both gas compression and cell wall compression is also listed. Values listed in **Table 1** are presented graphically in **Figure 8** and **Figure 9.** The compliance factor is plotted versus compressibility factor for 20 percent compression and 30 percent compression in **Figure 8** and work data are plotted in **Figure 9.** The reader is invited to explore these graphs

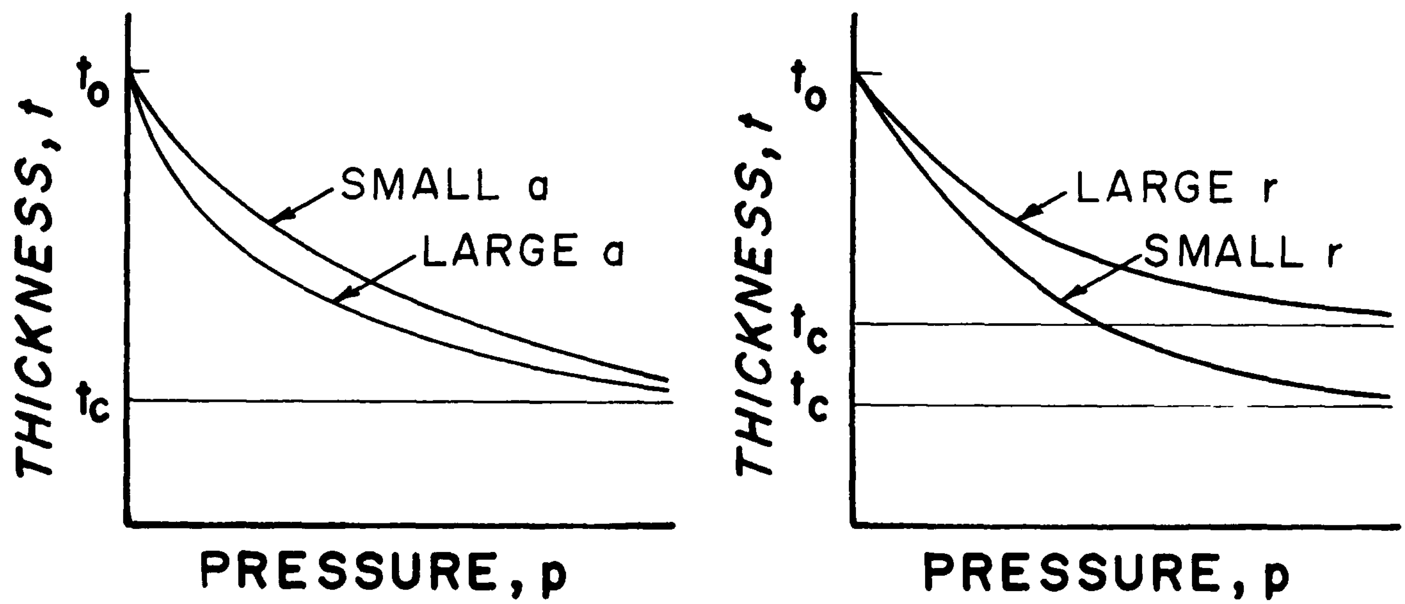

Figure 7.
Effects of two properties in the applied pressure/thickness relation. a) Compliance factor a; and, b) compressibility factor r.

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Table 1.
Characteristics of selected polyethylene foams. All values are normalized to one inch (2.54 cm) cube where a = compliance factor, r = compressibility factor, and w = work (inch-pound) for compression from 0 to 40 psi (276 kPa).

	a	r	w
1. Evazote	.0519	.222	22.0
2. Soft Pelite	.0414	.190	23.4
3. Medium Pelite	.0483	.394	26.5
4. Firm Pelite	.0292	.420	30.5
5. Aliplast 4E	.0427	.249	24.4
6. Aliplast 6A	.0317	.377	28.1
7. Med. Plastazote	.0443	.271	24.3
8. Firm Plastazote	.0472	.468	28.3
9. Med. Dermaplast	.0317	.237	26.8
10. Firm Dermaplast	.0514	.479	28.0

by first examining the effects of different compliance factors at a given compressibility factor. Note that the material with the larger compliance factor requires less pressure to compress the material by a given percentage. Next, examine the effects of different compressibility factors at a given compliance factor. Note that materials with a larger compressibility factor require more pressure to compress the material by a given percentage. Finally, note that certain combinations of compliance and compressibility factors require the same applied pressure to produce the same percent compression. A similar explanation may be followed regarding work relating to different combinations of compliance and compressibility factors, as shown in **Figure 9.**

In order to more clearly specify the nontime-related compression behavior of cellular foams, a comparison of the compression behavior of two polyethylene foams with the adiabatic compression behavior of air is presented in **Figure 10.** This graph suggests that under rapid loading

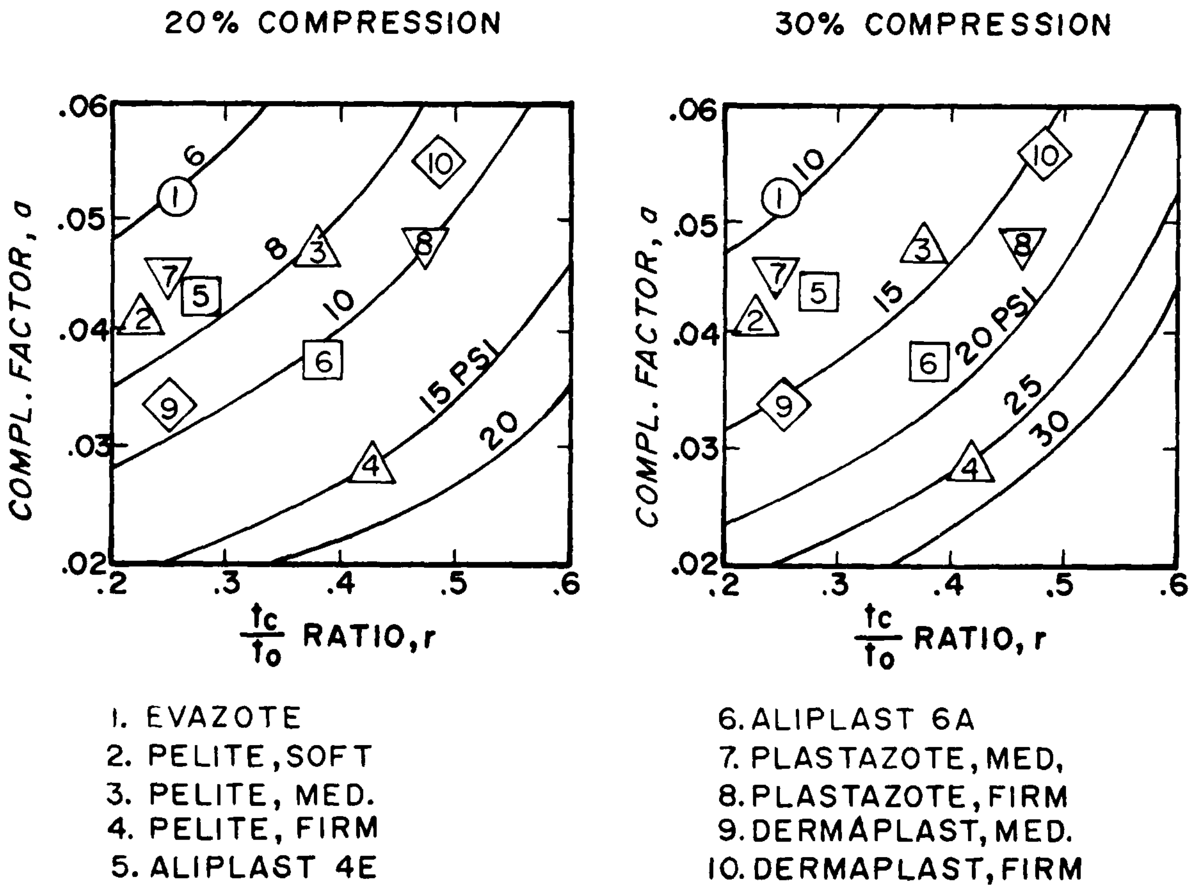

Figure 8.
Distribution of 10 different materials by compliance and compressibility factors showing pressure parameters required to compress the materials 20 and 30 percent.

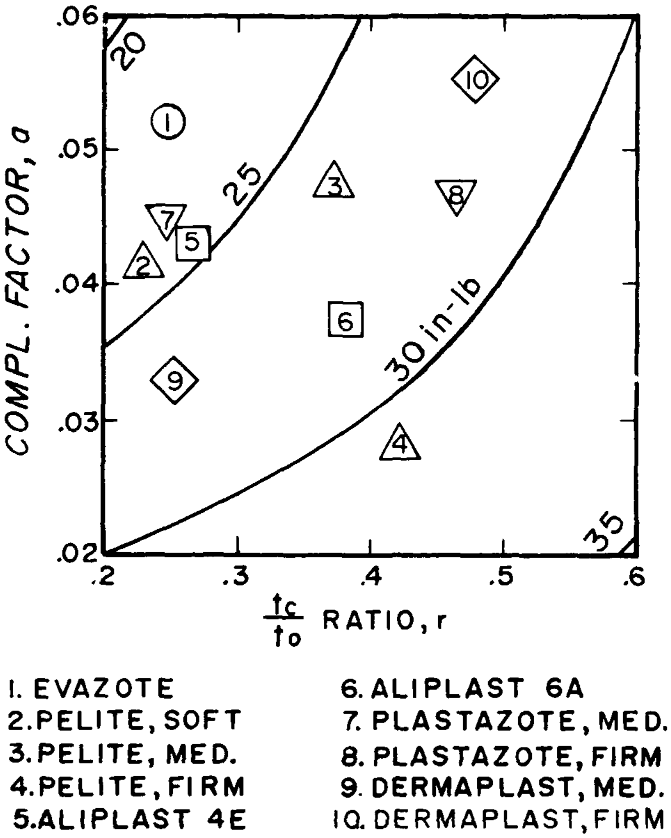

Figure 9.
Distribution of 10 different materials by compliance and compressibility factors showing work parameters (in·lb) required to compress a 1-inch (2.54 cm) cube of material with 40 psi (276 kPa) pressure.

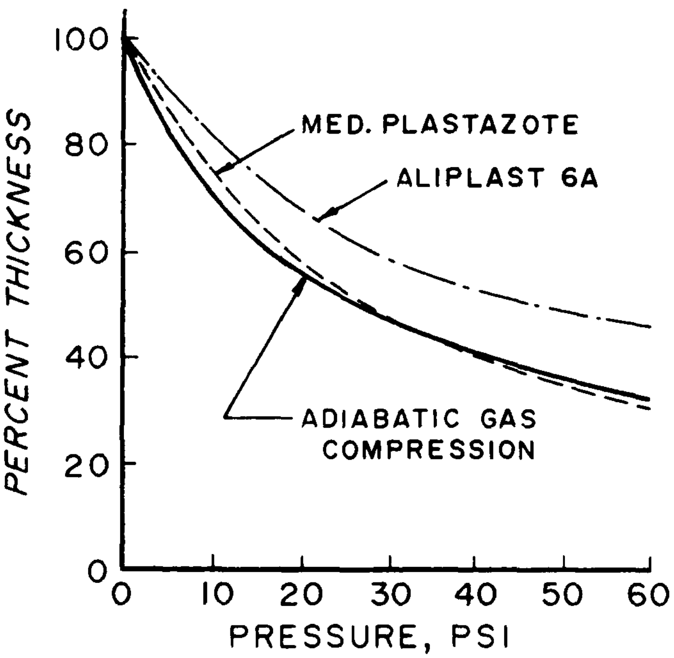

Figure 10.
Comparison of compression behavior of two polyethylene foams with adiabatic compression behavior of air.

the plastic foam materials tend to conform to the gas law. That is, the gas provides the major factor in the relation between pressure and thickness. Presumably, differences in the compression behavior of the foams and the gas are due to the resistance of the plastic foam matrix to elastic deformation. The difference in the compression behavior for Plastazote and Aliplast in **Figure 10** therefore indicates a dependence of elastic deformation on the value of the compliance factor, a. Work per one-inch (2.54 cm) cube of gas pressure in the compression of the gas from 0 to 40 psi (276 kPa) is 23.6 in·lb (2.6 J). This is less than for any of the foam materials indicating some degree of lost work. That is, some energy is not returned but is converted instead into heat and structural damage.

The abstractiveness of compliance and compressibility factors can be reduced through graphical illustrations which demonstrate their effects on pressure profiles. Picture a spherical surface being pushed into a plastic foam and then imagine the resulting pressure profile at that surface. To help visualize this setting, refer to **Figure 11** and **Figure 12.** To demonstrate the effects of different materials, the mathematical equations given earlier were used to calculate pressure profiles. In each case, a spherical surface of 3-inch radius (7.62 cm) is shown pushed 0.255 and 0.125 inches (0.57 and 0.32 cm) into plastic foams 0.375 and 0.50 inch (0.95 and 3.04 cm) thick. Aliplast 6A and Aliplast 4E, in **Figures 11** and **12** respectively, were selected only for illustrative purposes because of differing compliance and compressibility factors. For the illustration, the following factors were used: Aliplast 6A, a = 0.037 and r = 0.337; Aliplast 4E, a = 0.043 and r = 0.25. As anticipated, the resulting peak pressures are dependent upon the physical properties of the plastic foams, their thicknesses, and depth of compression. These diagrams demonstrate the critical role that choices in material and thickness play regarding interface pressures. Within the realistic limits presented in **Figures 11** and **12,** the peak pressures range from a minimum of 10 psi (69 kPa) to a maximum of 90 psi (620 kPa). The magnitudes of peak pressures under varying conditions infer several aspects to patient care. First, the relative sensitivity to depth of compression suggests that a small reduction in depth of compression may result in a substantial reduction in pressure. Second, choices in physical properties and thickness of the plastic foam can significantly affect the resulting pressures under like conditions. Third, cyclic loading under high pressure conditions will squeeze out gases more quickly from the cellular structure than under lower pressure, leading to reduced cushioning in a shorter interval of use. That is why foam plastics in this category tend to mold to the contour of the plantar surface during extended use. Conspicuous by its

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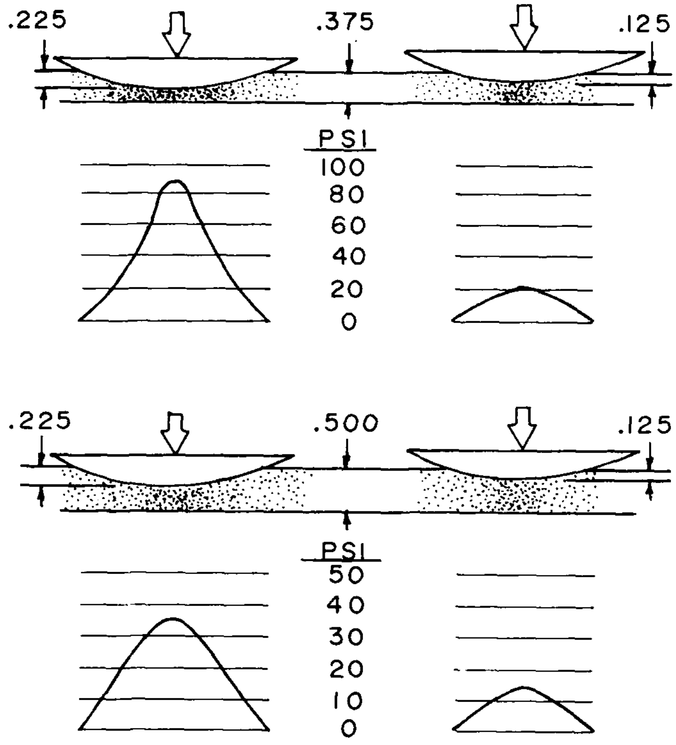

Figure 11.
Pressure profiles at the interface of a 3-inch (7.62 cm) radius spherical surface for two depths of compression for three-eighths and one-half inch (0.95 and 1.27 cm) thick Aliplast 6A.

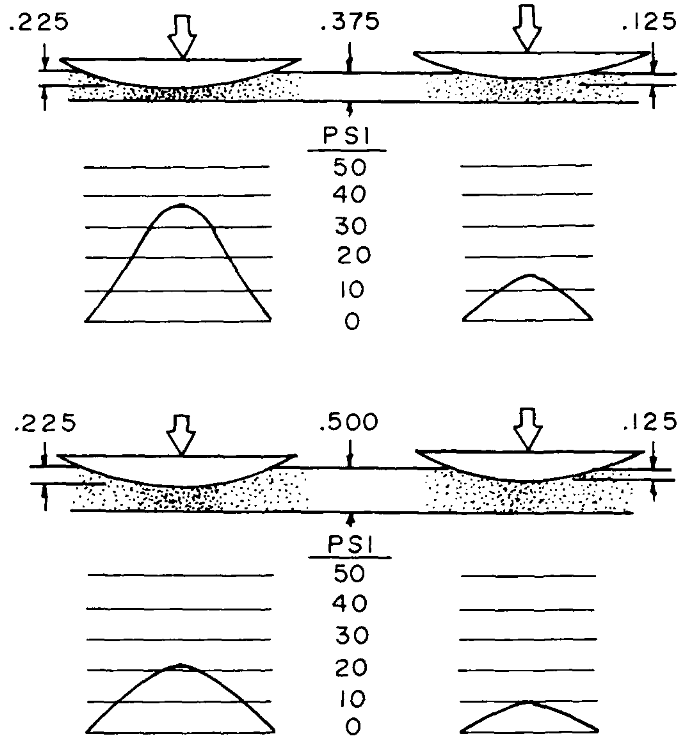

Figure 12.
Pressure profiles at the interface of a 3-inch (7.62 cm) radius spherical surface for two depths of compression for three-eighths and one-half inch (0.95 and 1.27 cm) thick Aliplast 4E.

absence is any reference to what pressures are tolerated by human plantar surfaces. These issues are beyond the scope of this presentation.

The relation between thickness and time as a result of a single loading cycle may be seen in **Figure 13.** Time-related properties were determined by a static load tester. The tester consisted of a pivoted beam, jaws for the test sample, a pointer, and a scale. The sample was placed in the jaws and a weight was placed at the proper location on the beam to represent 10, 20, or 30 psi (69, 138 or 207

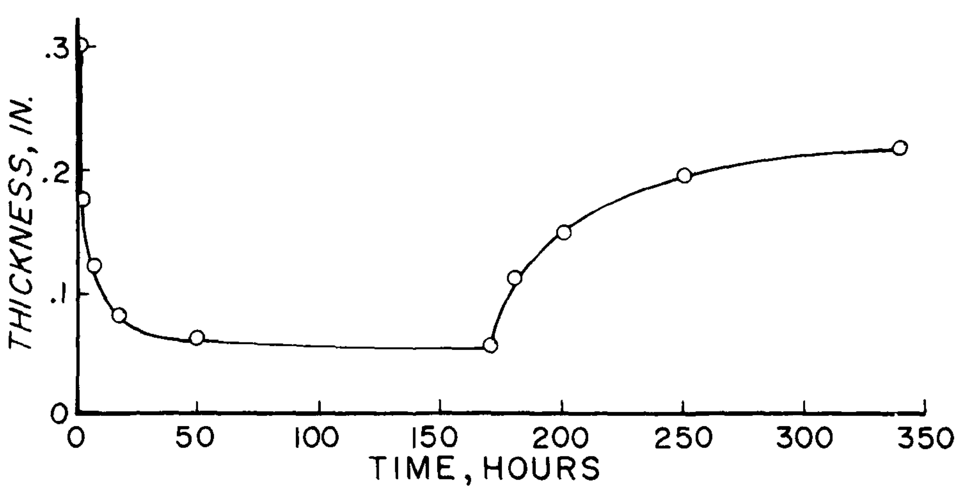

Figure 13.
Compression and recovery of medium density Plastazote. The static load was about 20 psi (138 kPa) applied for one week.

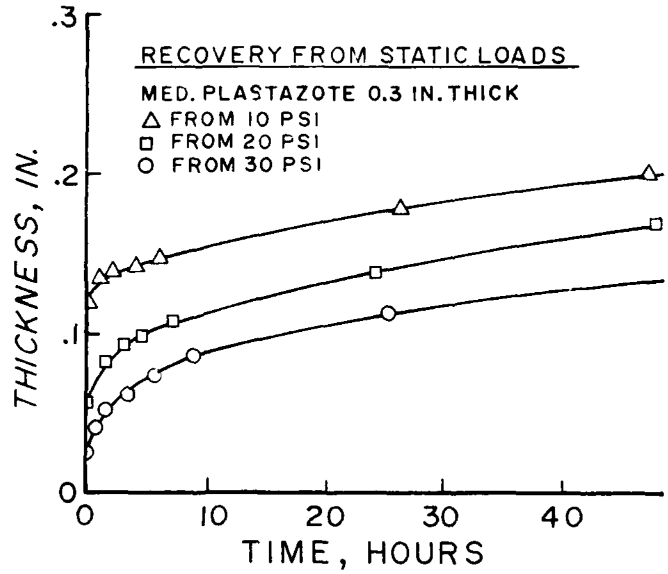

Figure 14.
Recovery of medium density Plastazote from different static loads applied for one week.

kPa) applied pressure on the sample. The thickness of the sample was displayed on the scale. Illustrated is the effect of a static load of about 20 psi applied for one week on a sample of medium density Plastazote 0.3 inch (0.76 cm) thick. The material reached equilibrium at about one-sixth of the original thickness in about 100 hours. When the load was removed, the thickness reached equilibrium at about 77 percent of original thickness in about 180 hours. Static load recovery data for Plastazote at loads of 10, 20, and 30 psi are shown in **Figure 14.** The recovery thickness depends on the initial compression as determined by the compressive load. The effect of cyclic sustained loading may be seen in **Figure 15.** In this test, a one-half inch (1.27 cm) thick sample of Aliplast 4E was cyclically loaded with approximately 30 psi. The load was applied in the morning and removed 8 hours later. Measurements of thickness were noted only at times of change of load. The thickness was recorded at 240 hours, approximately 130 hours after the conclusion of cyclic loading. The material had recovered to three-eighths of an inch (0.95 cm), or about 75 percent of original thickness. Cell window permeability accounts for this time-related behavior of the foam sample. As pointed out earlier, a sustained compressive load can squeeze out some of the gases from the cells. Upon removal of the load, the thickness will recover to varying extents depending upon the degree of mechanical integrity of the cellular structure.

In summary, these three examples of the time-related characteristics of closed cell plastic foams demonstrate the following: 1) during sustained loading, gas is forced out of the cells and more of the load is supported by elastic deformation of the cellular matrix; 2) when the load is removed, the potential energy stored in the plastic matrix

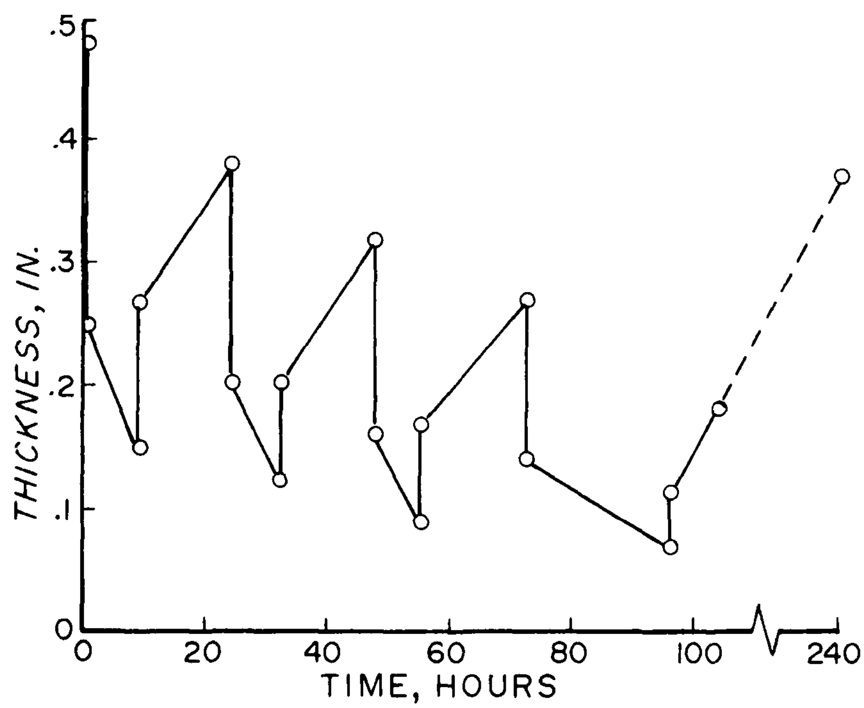

Figure 15.
Cyclic static loading of Aliplast 4E. Load was approximately 30 psi (207 kPa).

starts drawing air inward in order to refill the cells; and, 3) as the spring strength decreases with increasing thickness, the recovery rate decreases.

CONCLUSIONS

In the clinical setting, awareness of the previously discussed factors have allowed us to select ways to provide appropriate foot/insole interface materials and to maintain their use over prolonged periods of time. For example, many of the patients we encounter in our clinical study tend to obesity (i.e., 200 to 300 pounds, or 441 to 661 kg). In order to reduce the otherwise high plantar pressures, our results suggest the use of thicker interface materials. Moreover, an insole rotation procedure was instituted as a result of observed foam recovery properties. This procedure consisted of issuing several pairs of insoles for daily changes in order to permit insoles compressed during a day's use sufficient time to recover in thickness during the interval of nonuse.

In conclusion, the practical implications of these illustrations and comments are to increase the awareness of factors important in choices of available materials for applications such as insoles for footwear. Materials are available in a variety of cell structures, compliance, compressibility, and thickness. These foam characteristics should be considered in the treatment of patients of different weights and varied contours of the plantar surface. Our observations of the time-related and nontime-related phenomena suggest an avenue of improved care for the patient requiring insoles for footwear. Issuing thicker insoles offers a reduction of pressure concentration experienced by certain categories of patients. Moreover, issuing several pairs of insoles for daily rotation allows a recovery period during the nonwear interval to offer better cushioning during a day of wear.

GLOSSARY

Adiabatic. A change in volume or pressure without a change in thermal energy.

Bubbles. Referring to cells in a cellular foam.

Cell size. Indicating the relative size of cells or bubbles which compose a cellular foam.

Cell wall. Structural component separating two cells; also known as cell window.

Cellular foam. A mass of bubbles composed of a plastic and a gas phase.

Closed cell. Discrete cell through which gases do not pass freely.

Compliance factor (a). Characterizes the rapidity at which thickness changes with load in compression.

Compressibility factor (r). Characterizes the fraction of the unloaded thickness at which a limit is approached in compression.

Compressive load. Normal force per unit area in compression.

Density. Mass of polymer material per unit volume of the material.

Hysteresis. Failure of related phenomena to keep pace with each other (i.e., the delay in recovery of thickness upon removal of load under conditions of rapid loading).

Nonreticulated. Structure with intact walls separating the cells.

Nontime-related. Taking place under rapid cyclic loading conditions as may occur during a gait cycle.

Open cell. Discrete cell with open cell window through which gases may pass freely.

Potential energy. Energy stored in the cellular matrix as a result of compression.

Pressure (p). Normal force per unit area.

Reticulated. Structure with openings in cell walls.

Rib. Structural component of the cell occurring at the line of cell intersections which is capable of resisting a compressive load; also known as strand.

Time-related. Occurring under sustained loading conditions as in static loading or extended cyclic loading.

Work (w). Addition of energy through compression (i.e., change in volume as a result of applied pressure).

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Sitting forces and wheelchair mechanics

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Abstract—The effects of back angle and leg height on sitting forces in a wheelchair were studied, using a force plate mounted on a wheelchair seat. Readings of both normal force (perpendicular to the seat) and shear force were measured while the chair's back angle and footrest height were changed. Pressure under the ischial tuberosities was also measured during the footrest height adjustments. Five normal subjects sat directly on the plate as well as upon ROHO and Jay cushions placed on the force plate. Returning the back to the upright position after a recline caused the normal force (± SD) to increase 5.4 ± 2.5, 9.5 ± 4.0, and 10.0 ± 2.3 kg for the hard surface, Jay cushion, and ROHO cushion respectively, while shear at the plate increased to 5.1 ± 2.2, 11.6 ± 2.6, and 12.3 ± 2.7 kg for the hard surface, Jay cushion, and ROHO cushion respectively. Leaning forward (away from the back) caused all the forces to return to measurements close to the starting values. The results suggest that the wheelchair user should momentarily lean forward after a recline to reduce undesired forces. If a cushion with firm thigh support is used, ischial tuberosity pressure can be reduced by lowering the leg height as much as possible, which causes a levering action by lifting the pelvis.

Key words: *cushions, decubitus ulcer, force plate, pressure sore, seating pressure, shear force, wheelchair.*

INTRODUCTION

Wheelchairs have adjustable supports that are traditionally positioned for occupant comfort or by common sense. Little has been done to determine quantitatively the effect of support position on seat forces. This is especially true for shear force. Although back displacement has been used as a criterion to determine the suitability of different recline mechanisms (4), the resultant forces have not been measured. The interest in shear stems from the observation that shear force increases the possibility of inducing a decubitus ulcer. High skin shear levels have been found to lessen the normal force (downward force perpendicular to the supporting surface) needed to occlude underlying blood flow to one-half that needed when no shear is present (1). It is therefore important to reduce shear as much as possible. Another adjustable parameter which has not been fully studied is leg position. Changing footrest height may affect pressure on the ischial tuberosities and reduce the risk of ulcer formation. This study examines both the shear and normal forces under test subjects while an experimental wheelchair's back angle is changed, and the pressure under the ischial tuberosities as leg position is altered.

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METHODS

Experiments were conducted on an Everest & Jennings Premier model powered reclining wheelchair (Everest & Jennings, Los Angeles, CA). It had footrests that would elevate as the back reclined, a commonly-used wheelchair feature for persons with high-level quadriplegia. The armrests were removed for these experiments. Back angle was measured in degrees from vertical, as shown in **Figure 1**. Leg height was measured by thigh angle in degrees from horizontal with the positive direction being knees elevated.

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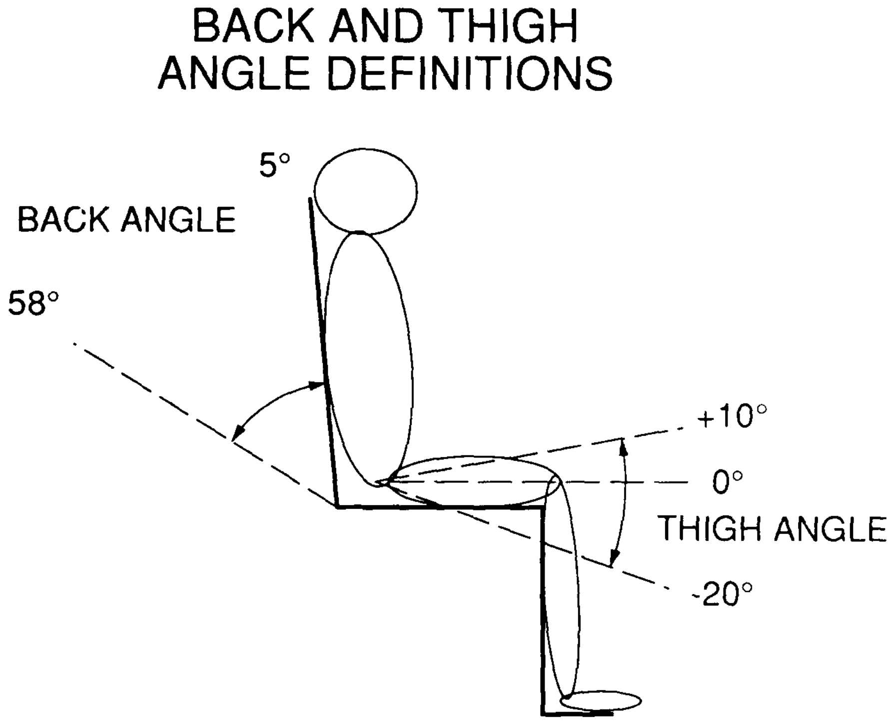

Figure 1.
Experimental setup for determining back and thigh angles.

The measurement was obtained using a board placed on the subject's lap.

An AMTI force plate (AMTI, Newton, MA) replaced the standard wheelchair seat. It was used to measure normal force, shear force, and the moment around an axis extending laterally from the center of the plate (**Figure 2**). The moment was divided by the normal force to determine the anterior-posterior position of the center of normal sitting force. Position readings closer to the rear of the plate indicate a greater percentage of force on the ischial tuberosities, while readings nearer the front indicate a greater force percentage on the thighs. As shown in **Figure 2**, normal force is positive in a downward direction, while shear force and anterior-posterior position are both positive in the anterior direction. Zero position is at the center of the plate.

Due to the thin construction of the plate, it was difficult to obtain the correct normal force readings along the edges. A modification done to improve accuracy is shown in **Figure 2**. First, a 3-mm thick sheet of hard plastic was fastened to the surface of the plate, which was cut back 4 cm from the edges of the plate. A metal sheet, of the same thickness as the plastic sheet, but with the same outer dimensions as the plate surface, was then fastened on top of the plastic sheet. This would redirect normal forces from the edge to the center of the plate. After modification, normal force readings of a 25 kg weight placed over any portion of the plate's surface differed by less than 5 percent.

Data were collected using an Apple IIe (Apple Computer, Cupertino, CA) with an Applied Engineering 12-bit A/D converter (Applied Engineering, Carrollton, TX) through a custom interface box. Strain gauge bridge circuits in the force plate were excited with 4.5 volts. All resultant analog voltage output signals were run through a two-pole low-pass Butterworth filter with a cutoff frequency of 3.5 Hz before being digitized. Readings were calculated by averaging samples taken at 10 Hz for 5 seconds.

Two separate test groups of five subjects each were used for the back recline and leg height experiments. Informed consent was obtained from each subject. All subjects were apparently normal healthy adults, two female and eight male, ages 24 to 50. The average weight of the test subjects was 73.8 kg for the back recline study, and 79.9 kg for the leg height study.

For the back position study, test subjects were placed in the wheelchair and the footrest height was adjusted so that the tops of their thighs were horizontal. Instruction was given to relax and not voluntarily change body position during the experiment. The wheelchair back was

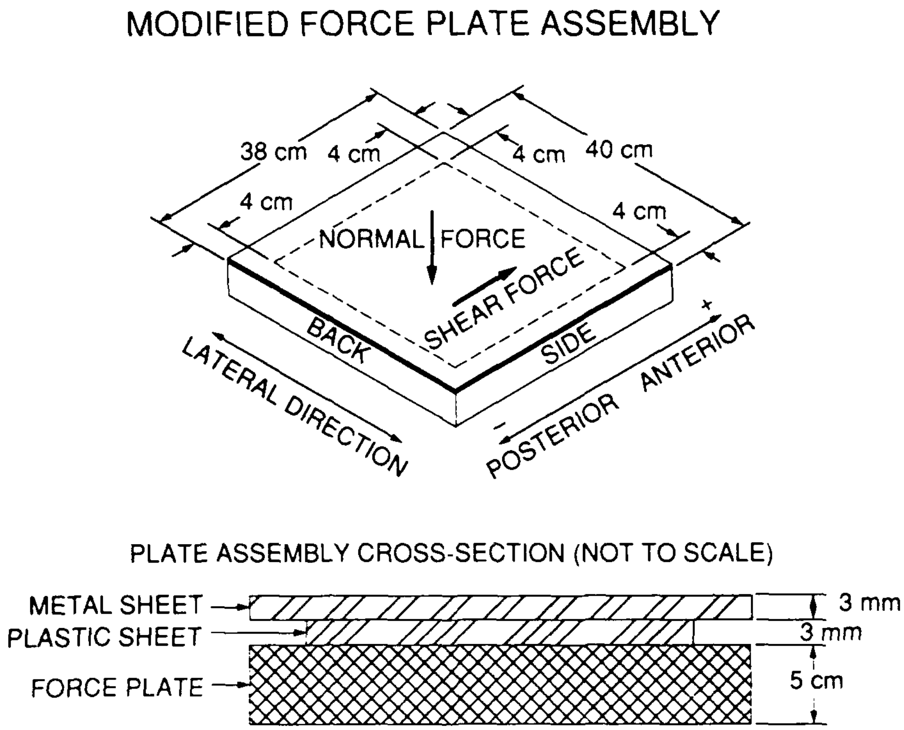

Figure 2.
Modified force plate with axes and force directions defined.

reclined at increasing angles beginning in a full upright position of 5 degrees from vertical and reclined in 5-degree intervals until a full recline position of 58 degrees was obtained. It then was returned to the upright position using the same intervals. Readings were taken at each interval. Two more recline cycles were then done, stopping to measure only at the 5- and 58-degree positions. Finally, the subject was asked to momentarily lean forward to remove any shears that might have accumulated over the back, and a final reading was taken. None of the intermediate positions were held for more than 30 seconds while the test was in progress.

The effects of leg position were also investigated. With the wheelchair back in full upright position, measurements were taken with the feet dangling, and then with the thighs at −10, 0, and +10 degrees (**Figure 1**). Thigh angle was changed by elevating the feet. Pressure under the ischial tuberosities was monitored with a Scimedics pressure evaluator (Scimedics, Inc., Anaheim, CA) at each thigh angle. Pressure readings were recorded by hand at each position.

Each subject was tested separately on ROHO cushions (ROHO Inc., Belleville, IL) and Jay cushions (Jay Medical Ltd., Boulder, CO), and also on the force plate with no cushions. When sitting directly on the force plate, a block of wood 1.4 cm thick was placed under the plate to keep the subjects at the same relative height in the chair. The order in which the cushions were used by each subject was randomized.

RESULTS

Table 1 and **Table 2** give the means and standard deviations (SD) of forces on the cushions as a function of the back angle. Normal forces were standardized by dividing each subject's normal force reading by their body weight, averaging these values at each test position, and then multiplying by the average body weight of all subjects. This technique was chosen to give results in kilograms for a representative subject with an average body weight of 73.8 kg. **Figure 3** and **Figure 4** show the results of normal and shear force measurements over the complete range of recline. Both the normal and shear force data were adjusted so that the graphs show change starting from zero. All the force values are given in kilograms equivalent at normal gravity in order to be easily compared to body weight.

The measured force and shear changes from the ROHO and Jay cushions were indistinguishable from each other, while the hard surface changes showed less force

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Table 1.
Shear and normal forces at various back angles.

		Normal force (kg)			*Shear force (kg)*		
		Hard	**ROHO**	**Jay**	**Hard**	**ROHO**	**Jay**
Initial upright position	ave	51.8	60.3	55.8	2.76	5.30	3.95
	SD	1.18	1.54	1.54	1.98	2.03	1.34
Full recline position	ave	37.4	43.7	41.1	−4.40	−1.08	−3.22
	SD	3.61	4.23	4.31	1.61	0.839	1.58
Return to upright position	ave	57.4	70.4	65.2	7.91	17.69	15.5
	SD	1.65	1.85	3.44	2.19	4.38	3.13
Upright position after lean	ave	51.6	60.0	57.5	3.42	8.17	7.57
	SD	1.42	2.03	4.09	1.63	6.79	1.12

buildup as the back was being returned to a vertical position. All subjects noted a "squeezed" feeling corresponding to this measured seat force buildup as they reached the full upright position. This buildup of forces was largely eliminated by the forward lean of the subject away from the back, which released force buildup along the back and returned the forces to very near the starting values.

Moving the chair back to full recline from full upright position reduced the normal seat force by 14.5 and 16.8 kg and changed shear by 6.3 to 7.2 kg on all surfaces. In all cases, the direction of shear reversed. The major differences between surface types showed up only when the back was returned to the upright position. At return to full upright position, the ROHO and Jay cushions showed a normal force ($\pm$ SD) increase of 10.0 $\pm$ 2.3 and 9.5 $\pm$ 4.0 kg, respectively, from initial readings, while the hard

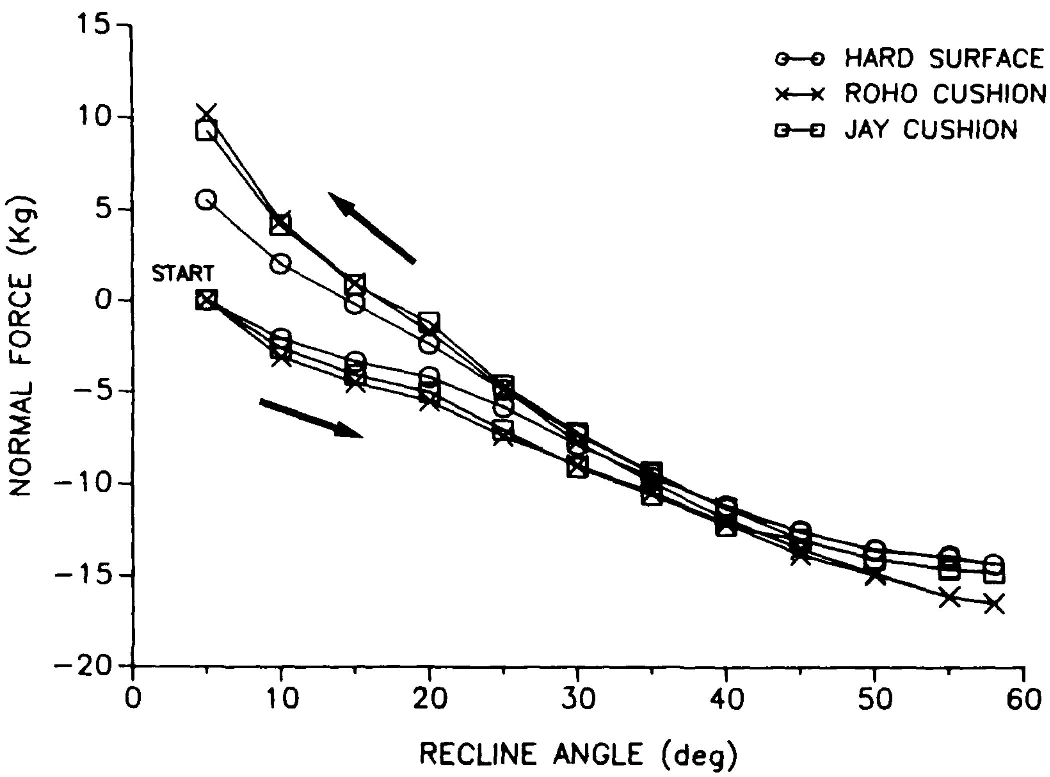

Figure 3.
Normal force versus recline angle of the back (degrees from vertical). The data show the change in force with the starting value defined as zero. Arrows indicate the direction of movement.

Table 2.
Ischial tuberosity pressure, anterior-posterior position and normal force at dif ent thigh angles.

		Pressure (mm Hg)			Position (cm)			Force (kg)		
		Hard	ROHO	Jay	Hard	ROHO	Jay	Hard	ROHO	Jay
Legs dangling	ave	157.6	68.4	59.8	2.30	1.94	3.10	75.9	79.8	76.9
	SD	26.1	4.39	11.3	1.18	1.34	1.35	3.06	4.06	3.27
Legs at −10 degrees	ave	211.6	65.8	76.6	−4.36	−0.640	−1.84	60.0	67.8	63.1
	SD	45.0	4.02	7.20	1.06	1.34	1.28	2.71	3.60	2.27
Legs at 0 degrees	ave	251.2	70.6	85.2	−6.62	−2.17	−3.92	55.5	64.7	59.4
	SD	35.4	3.85	7.85	0.370	1.29	0.963	2.32	3.91	2.91
Legs at +10 degrees	ave	256.2	73.4	86.6	−7.65	−3.40	−5.48	53.4	62.0	57.0
	SD	27.1	4.98	7.89	0.493	1.26	0.719	1.79	3.00	2.13

surface showed an increase of only 5.4 ± 2.5 kg. The ROHO and Jay cushions showed shear force changes from rest of 12.3 ± 2.7 and 11.6 ± 2.6 kg, respectively, while shear on the hard surface changed to 5.1 ± 2.2 kg from a resting position. The data accumulated from the two subsequent recline cycles revealed no force buildup with any of the three surfaces.

The results of the leg height study are shown in **Figure 5** and **Figure 6**, as well as in **Table 2**. It can be seen that as the legs were elevated, the normal force was reduced, while the center of force was shifted backward. This was due to the thighs being lifted off their supporting surfaces and their weight being transferred to the feet. The rearward weight shift was somewhat linear with the ROHO cushion, but with the hard surface and Jay cushion it shifted more quickly as the legs were raised from a dangling posi-

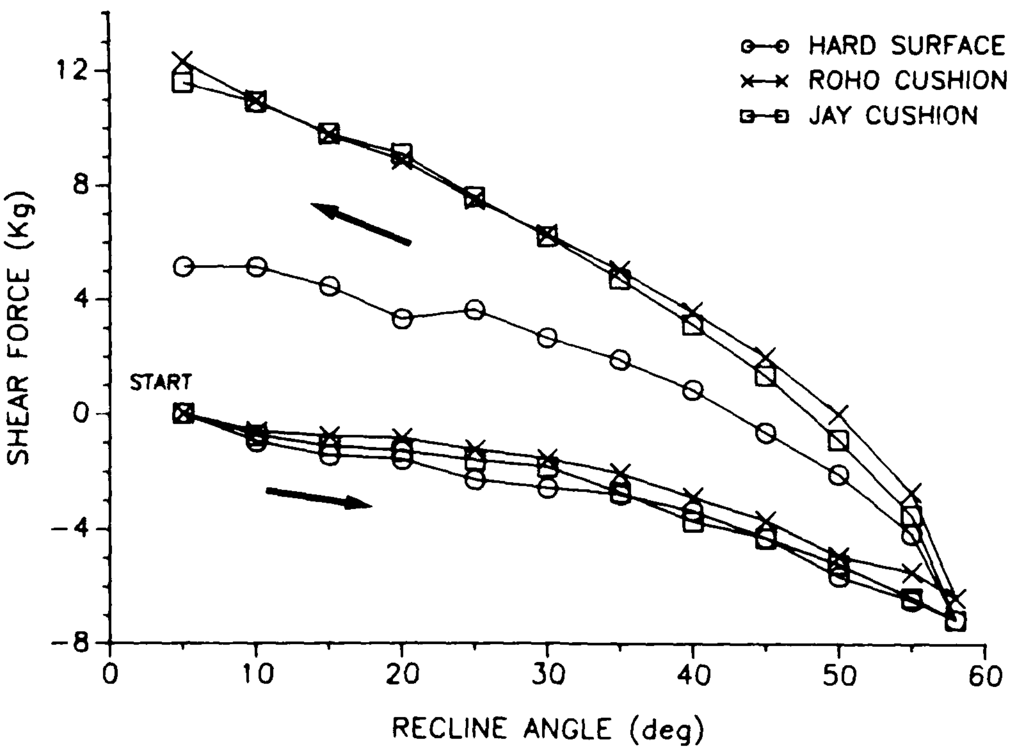

Figure 4.
Shear force versus recline angle of the back (degrees from vertical). The data show the change in shear with the starting value defined as zero. Arrows indicate the direction of movement.

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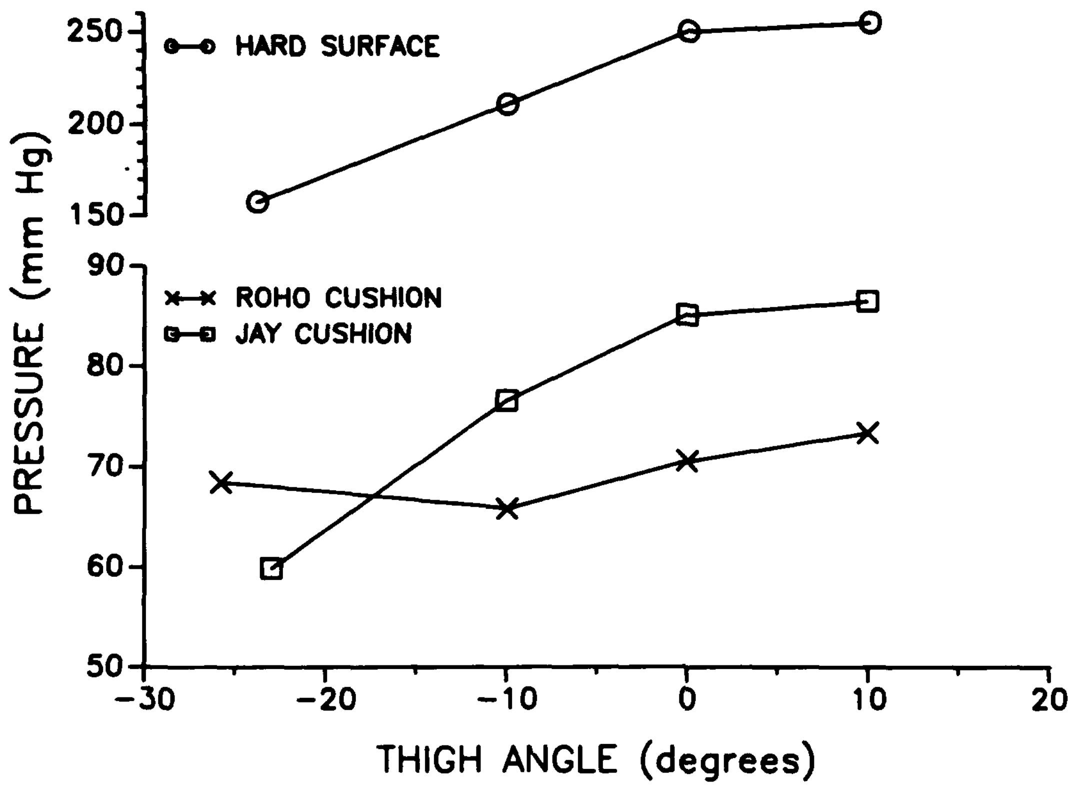

Figure 5.
Pressure on the ischial tuberosities as the legs were raised. The left-most point of each curve represents data taken with the legs dangling.

tion to −10 degrees. As the thighs were elevated from −10 degrees, the rates of weight shift approached each other. Pressures on the hard surface were very high and greater than with either cushion. As the legs were elevated, the pressure on the hard surface increased from 150 mmHg to 250 mmHg. Leg elevation while on the ROHO cushion produced a small pressure change from 68 to 73 mmHg, while on the Jay cushion a greater change occurred, from 60 to 87 mmHg. The thigh angle also varied with surface when the feet were dangling. The steepest angle was produced by the ROHO cushion, followed by the hard surface, with the Jay cushion allowing the thigh to drop the least amount.

DISCUSSION

In this study we have examined the changes in shear and normal seat forces brought about by back support and footrest height adjustment on a wheelchair. Ideally, knowledge of both the localized pressures and shears acting on the skin are desired. The system developed by Bennett and associates for measuring localized shear, pressure, and blood flow worked only with skin pressed against a hard surface (2). No practical device exists today that can accurately measure localized shear when used with soft cushions. The exact level of localized shear is unknown, but may follow the overall shear as indicated by the force plate.

The differences in initial normal forces can be explained by the different thigh support properties of the three surfaces. Lowering the feet transfers part of thigh and leg weight from the feet to the force plate, resulting in higher normal total force readings, but lower pressure over the ischial tuberosities. This is particularly true for the Jay cushion and hard surface.

The reclining of the wheelchair back results in a reduction of normal force, which helps alleviate tissue pressure. It also causes a reversal of shear force from forward to rearward. This shift in shear direction could change internal stress patterns and transfer some load to other tissue areas.

The forces encountered during the back elevation phase changed significantly as the full upright position was approached. Raising the chair's back from full recline to full upright position increased normal force by nearly 27 kg and added about 9 kg to shear force on the cushions. The increased forces, especially the shear forces, were found to be uncomfortable for even short periods, and could

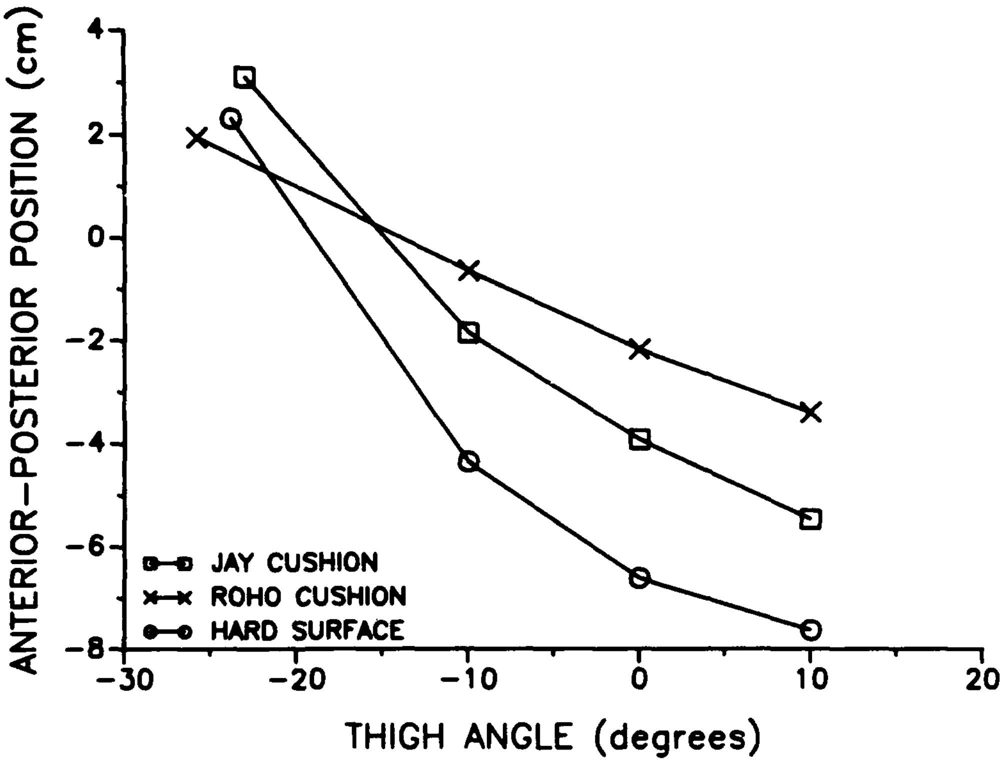

Figure 6.
Anterior-posterior position of sitting normal force versus thigh angle as the legs were raised. The left-most point of each curve represents data taken with the legs dangling.

be potentially harmful if sustained in patients with minimal sensation. This situation is caused by the sliding of the skin along the chair's back during the recline phase. A simple solution to eliminate the increased force is to momentarily lean the wheelchair occupant forward to remove contact with the back after each return to the upright position. This reduced shear force by more than a factor of two and reduced normal force by 10 percent, returning it to its pre-recline value. Other recline systems, such as a "tilt in space" wheelchair or the four-bar linkage design by Warren and associates (4), would be expected to yield different results due to differing chair back movement patterns in relation to the chair seat.

The problem of disabled people shifting downwards in their chairs was not reproduced during repeated reclines. As testing was performed upon able-bodied subjects, unconscious postural muscle activity might have played a part in the lack of sliding. Another factor may be the length of time that the subjects were held in full recline. As mentioned earlier, this time was approximately 30 seconds.

Some interesting points can be made concerning the leg height experiment. When the thighs were in the zero-degree position, the ROHO cushion showed 15 mmHg less pressure compared to the Jay cushion; but with the legs dangling, the Jay cushion produced the least pressure by 9 mmHg. The ischial pressure while on the ROHO is relatively independent of leg height, while a subject sitting on a Jay cushion can change the pressure through leg height changes. Bush (3), working with hard sitting surfaces, also reported this effect.

The tendency of the tuberosity pressure to drop while support is removed from the feet could be enhanced by shifting the thigh's pivot point rearward. With firm support surfaces such as that provided by the Jay cushion, this pivot point is on the front edge of the cushion. If the top of the cushion were to be shaved down near the front, and an elevation built up closer to the cushion's center, the extra thigh weight ahead of this new pivot point would cause a greater lever action for lifting the tuberosities. Although with the ROHO the pivot point may be nearer the tuberosities, the nature of the cushion causes it to apply constant pressure to the skin regardless of depth into the cushion and no levering action occurs.

It should be noted that it is not practical to leave the feet totally without support. However, if most of the leg weight were supported by the cushion, with only enough weight placed on the feet to keep the heels on the footrests,

the same effect would be produced. Wheelchair occupants might then, in fact, need only rock forward for pressure relief. This would require further study to verify.

CONCLUSION

Two major phenomena are demonstrated in this study. First, having the user lean forward after a wheelchair back recline will greatly reduce undesired force. Secondly, wheelchair cushions with firm material under the thighs will facilitate reduction in ischial tuberosity pressure when the leg height is lowered as much as possible. Implementation of these two findings may help to reduce tissue damage in wheelchair-bound individuals.

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Audio-visual consonant recognition with the 3M/House cochlear implant

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Abstract—Eight experienced 3M/House cochlear implant users' consonant recognition was evaluated with videotaped vowel-consonant vowel lists presented in auditory implant only (A), visual (V), and auditory-visual (AV) conditions. All subjects' scores were better than chance. Results revealed that the AV scores were significantly better than the V scores, which were better than the A scores. Sequential Information Analysis of the consonant errors revealed that different features were transmitted better in each condition. Sonorant and voicing features were transmitted well for the A condition, but features related to high-frequency and place cues were not. Place features were transmitted best in the V condition, but acoustic features were not. Both place and acoustic features were transmitted in the AV condition, but they were influenced most by visual cues.

Key words: *auditory-visual, cochlear implants, consonant recognition, 3M/House.*

INTRODUCTION

The single-channel 3M/House cochlear implant (6,12) has been used in over 800 profoundly deaf patients. Although several studies have examined speech perception by cochlear implant users, only a few have reported consonant recognition data for the 3M/House device in auditory implant only (A), visual (V), and auditory-visual (AV) conditions (5,8,10,15). Studies that have evaluated the 3M/House implant have generally found that the subjects' speech recognition abilities are poor. However, they have usually used only a few subjects tested in the auditory-only condition with various stimuli and contexts, and assumed that most 3M/House users perform poorly with their implants.

This study does not argue the merits of the 3M/House device, but provides additional information to the limited existing database for consonant recognition with this implant. Indeed, few studies have actually looked at A, V, and AV consonant recognition in a controlled fashion (i.e., using videotaped stimuli instead of live-voice face-to-face presentations which vary too much to provide a constant stimulus) with any of the cochlear implant devices that are available. Even fewer studies have analyzed the subjects' consonant errors and presented information about perceptual features obtained from Sequential Information Analyses (SINFA). This study evaluated closed-set consonant recognition by experienced users of the 3M/House device.

METHOD

Subjects were eight postlingually, profoundly deaf persons (7 women and 1 man) between 23 and 60 (M = 42) years of age who had used their 3M/House implants for at least two years. They represented a fair cross section of experienced users of the 3M/House implant, were in good general health, had normal or normally corrected

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visual acuity (20/40 using a Snellen vision chart), and intelligible speech.

The subjects were volunteers who were available, agreed to participate in testing from a group of about 25 patients meeting the selection criteria, and who were within driving distance to Los Angeles, CA. They were selected from the files of the Otologic Medical Group in Los Angeles and also are to be part of a larger study to be conducted later. They had not previously received the particular stimuli used in this study. Unfortunately, few preoperative speech perception data (and no auditory-only speech recognition scores) were available in the subjects' files. Generally, the subjects met the conditions for implantation with this device (i.e., no speech recognition with the use of traditional hearing aids prior to implantation). Although performance on the stimuli used in this study was not assessed preimplant, based on our experience with other users of this device, these stimuli would probably have been too difficult for them preimplant. The subjects' files revealed that their performance on the House Ear Institute Environmental Sounds tests was 75 percent or better.

Three videotape-recorded General American male-talker 60-item vowel-consonant-vowel (VCV) lists consisting of five random repetitions of each of 12 consonants (/p,b,k,g,ʃ,s,f,v,r,l,m,n/) in an /ʌCʌ/ context were presented via a videotape playback in A, V, and AV conditions. All items were produced in a natural fashion without facial exaggeration. The visual stimuli were close-up face and neck shots. All stimuli were judged to be equally intelligible by three normal-hearing listeners in the A and AV conditions. Stimuli were routed from the videotape playback through a mixer, amplifier, and attenuator, and directly connected to the microphone input of each subject's cochlear implant processor. This procedure bypassed the frequency characteristics of the microphone, eliminated body-baffle effects, and may have provided the subjects with an unfamiliar signal spectrum, but was used to control as much of the variability associated with the signal input as possible. This procedure is similar to the direct-connect condition used by Rosen, Walliker, Brimacombe, and Edgerton (13). Stimuli in the V and AV conditions were directed to 48.5 cm diagonal color video monitor.

Each subject was tested individually in a quiet, well-lighted room and received one stimulus list per presentation mode. Order of stimulus and presentation modes was randomized for each subject. The signal level at the input of the signal processor was adjusted to 3 mV peak-to-peak (∼79.5 dB SPL peak). Each subject adjusted the processor volume to a most comfortable level for connected discourse. Subjects received oral and written instructions, and a practice list of the 12 consonants at the beginning of each session. The list of the 12 possible consonants (closed-set) was available to the subjects throughout testing. Subjects' responses were phonetically transcribed and confirmed by the subject before progressing to the next item; guessing was encouraged and no feedback was provided.

RESULTS AND DISCUSSION

Figure 1 shows individual subject data for all conditions. Confidence intervals were calculated using the normal approximation to the binomial distribution to determine a chance score for the 60 items (5 repetitions $\times$ 12 consonants) of each list on this test; all scores were better than the 18 percent level required to be significantly above chance at the upper 99 percent confidence limit.

Wilcoxon matched-pairs signed-ranks tests for small samples (14) revealed that the subjects' VCV scores in the AV condition (M = 91 percent) were significantly better than in either the A (M = 47 percent) or V (M = 60 percent) conditions (both $T = 0$, $p \leq 0.01$). Performance in the AV condition was good, ranging from 72 to 100 percent correct as compared to 35 to 60 percent and 48 to 93 percent for the A and V conditions, respectively. All subjects' scores improved in the AV condition, even those for Subject #7 who has exceptionally good speechreading skills. The subjects' scores in the V condition were significantly better than those in the A condition ($T = 1$, $p \leq 0.01$).

The subjects' consonant errors were pooled and converted to confusion matrices for the A, V, and AV conditions as shown in **Table 1, Table 2**, and **Table 3**, respectively. The matrix for each condition was submitted to a Sequential Information Analysis (SINFA) (18), to determine if different types and amounts of information were transmitted by selected *a priori* features. The features specified for this analysis are provided in **Table 4**.

The SINFA program partials out the effects of one feature on another by first estimating the unconditional transmitted information for each feature in the feature system according to the percentage of information transmitted. SINFA then proceeds through a series of iterations in which the feature with the highest percentage of information transmitted is revealed for iteration number one (No. 1). In iteration No. 2, the feature identified in iteration No. 1 is held constant or partialed out, and the conditional transmitted information for the remaining features is determined (i.e., this transmitted information is independent of that for the feature identified in iteration No. 1). The feature having the highest percentage of transmitted

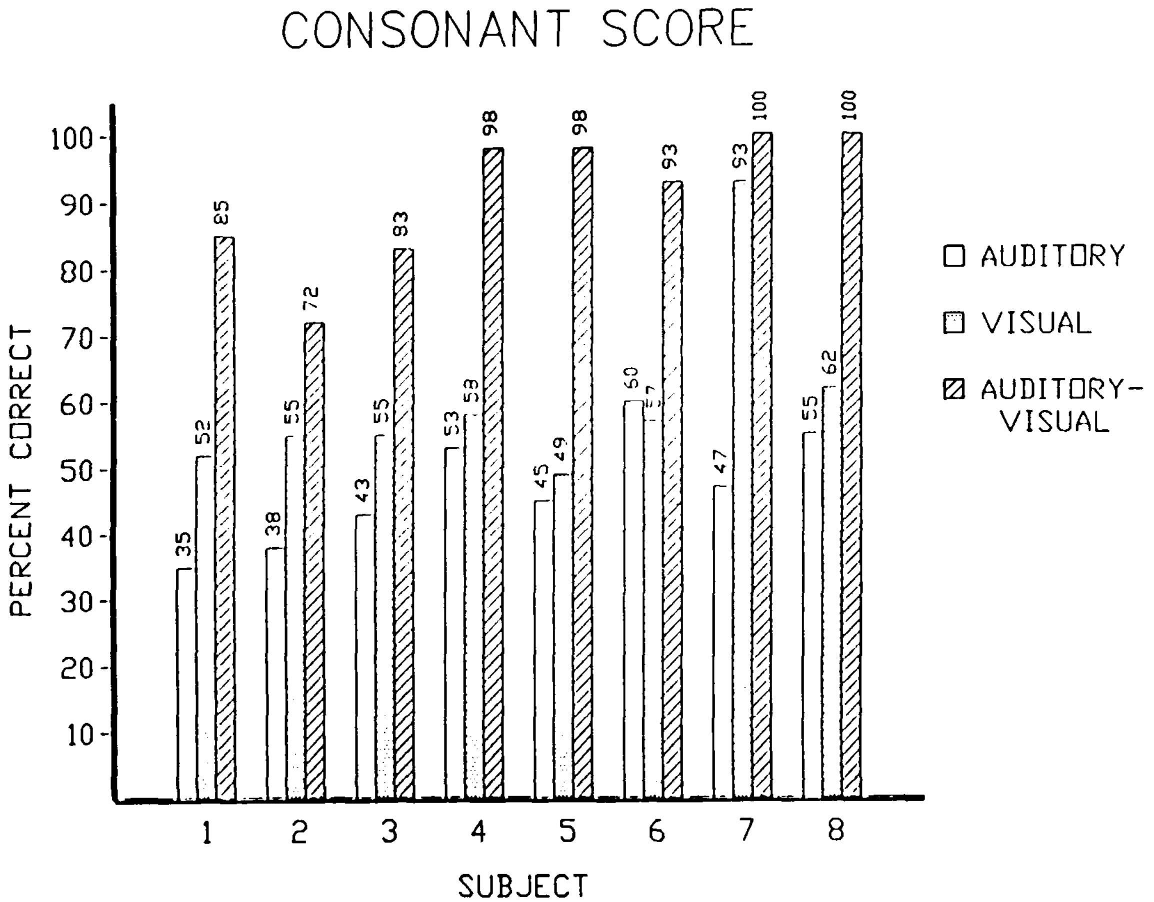

Figure 1.
Individual percent correct consonant recognition scores for the eight cochlear implant users in the auditory, visual, and auditory-visual modes (chance at 99 percent confidence interval > 18 percent).

information in iteration No. 2 is then identified and is held constant for iteration No. 3, and so on.

Table 5 summarizes the SINFA results for the A condition. The stimuli feature information computed for each feature in the system is presented in the first column. The conditional transmitted information in bits for each stimulus for the features is in the second column. The percentage of conditional information transmitted is in column three. SINFA also indicates when redundant information is associated with the features, and determines when remaining features in the system become equivalent. Features depicted in the later iterations are probably of negligible perceptual importance (18).

Table 6 summarizes the SINFA results for the three conditions and shows the features identified and the percent of (conditional) transmitted information for each iteration. The iteration number is included in parentheses. Features without an entry in certain columns were those having negligible contributions to the total transmitted information according to SINFA. Iterations No. 5 and 6 for the V, and No. 3, 6, and 7 for the AV conditions, were redundant as indicated by the same percentages of transmitted information.

Table 6 shows that with the exception of palatal, features transmitted well in the A condition related to sonorancy and voicing. A high degree of redundancy is inherent in these features, as all sonorants are voiced, and the nasals and liquids can also be classified as sonorants. These features involve low-frequency information that should be available through the 3M/House implant. However, high-frequency features (e.g., sibilant) and those related to place of articulation (e.g., front/back, bilabial, dental, alveolar, and velar) were not transmitted well by the implant. The subjects almost always recognized the palatals, /r/ and /ʃ/, correctly in the A condition. They may have been responding primarily to intensity and/or

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Table 1.
Pooled confusion matrix for auditory condition.

Stimulus \ Response	p	b	k	g	ʃ	s	f	v	r	l	m	n
p	19	1	8	3	2	2	5					
b	2	27		4	1		2	3		1		
k	18		12	1		5	3	1				
g	4	18	2	7		1	4	1	1	2		
ʃ			2		32	6						
s	2	1	2	4	2	15	14					
f	2	2	1	4		15	14	1		1		
v	1	8	1	5		1	4	19				1
r								1	36	3		
l									14	19	3	4
m		1				5		1	2	10	10	11
n						2			2	9	13	14

duration cues for /ʃ/, and to low frequency (e.g., F1 transition) cues for /r/. These two sounds are almost always identified correctly by these subjects in open-set contexts, in and out of test situations. As expected, features transmitted best in the V condition related to place of articulation (e.g., bilabial, dental, etc.), whereas acoustic features (e.g., voicing, sonorant, nasal, etc.) were not transmitted well. Sibilant was identified on iteration No. 5 for the V condition; however, the facial contortions involved in producing the sibilant sounds frequently make these (otherwise acoustic phonemes) highly visible and distinguishable from other phonemes (4).

Interestingly, the features identified in the AV condition resulted in a combination of both auditory and visual features. Sibilant was the first feature identified, but as just discussed, this may have been due more to visual than to acoustic cues. Other acoustic features identified in the A condition (i.e., voicing, sonorant, and nasal) were not identified until iteration No. 6. Thus, the feature information transmitted for these cochlear implant users seemed to be driven more by visual cues available through speechreading than by acoustic cues provided by their implants.

These results are in agreement with those from many other studies that have evaluated consonant perception in

Table 2.
Pooled confusion matrix for visual condition.

Stimulus \ Response	p	b	k	g	ʃ	s	f	v	r	l	m	n
p	16	17					1				6	
b	12	17									11	
k			15	22						1		2
g			12	18					1	3		6
ʃ					35	3						2
s					2	38						
f							27	13				
v							26	14				
r									40			
l				1						39		
m	13	18									9	
n			2	6	1	9				2		20

Table 3.
Pooled consonant confusion matrix for auditory-visual condition.

Stimulus \ Response	p	b	k	g	ʃ	s	f	v	r	l	m	n
p	39	1										
b	2	37						1				
k			37	3								
g			3	36								1
ʃ					37	3						
s					1	39						
f							32	8				
v				1			3	36				
r									40			
l										39		1
m	2	3									35	
n	1			2		2				1		34

Table 4.
Features specified for the sequential information analysis.

Phoneme	VOIC	FR/B	SIBL	SONR	NASL	LIQD	PLOS	FRIC	BILB	DENT	ALVR	PALT	VELR
p	0	0	0	0	0	0	1	0	1	0	0	0	0
b	1	0	0	0	0	0	1	0	1	0	0	0	0
k	0	1	0	0	0	0	1	0	0	0	0	0	1
g	1	1	0	0	0	0	1	0	0	0	0	0	1
ʃ	0	1	1	0	0	0	0	1	0	0	0	1	0
s	0	0	1	0	0	0	0	1	0	0	1	0	0
f	0	0	0	0	0	0	0	1	0	1	0	0	0
v	1	0	0	0	0	0	0	1	0	1	0	0	0
r	1	1	0	1	0	1	0	0	0	0	0	1	0
l	1	0	0	1	0	1	0	0	0	0	1	0	0
m	1	0	0	1	1	0	0	0	1	0	0	0	0
n	1	0	0	1	1	0	0	0	0	0	1	0	0

Table 5.
Sequential information analysis of consonant confusions in the auditory condition.

Iteration No. 1 Feature	Feature Inf.	Trans. Inf.	% Trans. Inf.
VOIC	0.980	0.514	0.525
FR/B	0.918	0.123	0.134
SIBL	0.650	0.193	0.297
SONR	0.918	0.708	0.771
NASL	0.650	0.240	0.369
LIQD	0.650	0.355	0.546
PLOS	0.918	0.335	0.365
FRIC	0.918	0.306	0.333
BILB	0.811	0.057	0.070
DENT	0.650	0.092	0.141
ALVR	0.811	0.078	0.096
PALT	0.650	0.330	0.508
VELR	0.650	0.029	0.044

(**Table 5** continues on the following two pages)

Table 5. (continued)

Iteration No. 2
Feature Held Constant: SONR

Feature	Cond. Feat. Inf.	Cond. Trans. Inf.	% Cond. Trans. Inf.
VOIC	0.628	0.257	0.409
FR/B	0.904	0.161	0.178
SIBL	0.536	0.141	0.263
NASL	0.322	0.091	0.282
LIQD	0.322	0.091	0.282
PLOS	0.666	0.173	0.260
FRIC	0.666	0.173	0.260
BILB	0.796	0.064	0.080
DENT	0.541	0.055	0.102
ALVR	0.687	0.041	0.060
PALT	0.628	0.322	0.513
VELR	0.538	0.011	0.021

Iteration No. 3
Features Held Constant: SONR, PALT

Feature	Cond. Feat. Inf.	Cond. Trans. Inf.	% Cond. Trans. Inf.
VOIC	0.566	0.212	0.374
FR/B	0.494	0.007	0.014
SIBL	0.339	0.026	0.078
NASL	0.194	0.023	0.121
LIQD	0.194	0.023	0.121
PLOS	0.573	0.124	0.217
FRIC	0.573	0.124	0.217
BILB	0.710	0.044	0.062
DENT	0.497	0.040	0.080
ALVR	0.551	0.029	0.052
VELR	0.494	0.007	0.014

A, V, and AV conditions by normal-hearing listeners, hearing-impaired hearing aid wearers, and single-channel cochlear implant users (1,3,7,9,10,11,13,16,17). Although direct predictions to conversational speech cannot be drawn from these results, the improvements in consonant scores in the AV condition suggest that the implant should help these subjects perform well in everyday speechreading situations having more content and contextual cues. The subjects' scores in the V condition generally reflected their average-to-excellent speechreading ability. Their performance in the AV condition was probably due to their ability to combine the auditory cues provided by the cochlear implant with visual cues and not just due to speechreading alone.

In summary, this study adds to the database on consonant recognition with cochlear implants and provides

Table 5. (continued)

Iteration No. 4
Features Held Constant: SONR, PALT, VOIC

Feature	Cond. Feat. Inf.	Cond. Trans. Inf.	% Cond. Trans. Inf.
FR/B	0.481	0.016	0.032
SIBL	0.268	0.011	0.039
NASL	0.194	0.023	0.121
LIQD	0.194	0.023	0.121
PLOS	0.554	0.156	0.281
FRIC	0.554	0.156	0.281
BILB	0.697	0.039	0.056
DENT	0.487	0.057	0.116
ALVR	0.477	0.011	0.023
VELR	0.481	0.016	0.032

Iteration No. 5
Features Held Constant: SONR, PALT, VOIC, PLOS/FRIC

Feature	Cond. Feat. Inf.	Cond. Trans. Inf.	% Cond. Trans. Inf.
FR/B	0.311	0.007	0.022
SIBL	0.158	0.000	0.000
NASL	0.194	0.023	0.121
LIQD	0.194	0.023	0.121
BILB	0.521	0.007	0.014
DENT	0.158	0.000	0.000
ALVR	0.367	0.001	0.002
VELR	0.311	0.007	0.022

consonant recognition data for A, V, and AV conditions using the 3M/House implant. Further, it provides findings about how much information was transmitted by each perceptual feature in each condition. Primary features for the A condition were voicing, nasality, and sonorant; those for the V condition related to place of articulation; and those for the AV condition were a combination of those for the A and V conditions plus sibilancy. The study shows which features were and were not transmitted well by this device. Visual features were transmitted better than auditory-only features, but subjects performed very well when both auditory and visual cues were available, even with the 3M/House device that is admittedly simpler than some other implants currently available. Assuming that the other devices are capable of providing more acoustic cues, patients using those devices may be expected to perform even better than patients using the 3M/House device.

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Table 6.
Summary of percentage of transferred information from sequential information analysis for auditory, visual, and auditory-visual conditions.

Feature	Auditory	Visual	Auditory-Visual
VOIC	(3) 0.374		(8) 0.605
FR/B		(6) 0.282	(6) 0.806
SIBL		(5) 0.714	(1) 0.957
SONR	(1) 0.771		(7) 0.751
NASL	(5) 0.121		(7) 0.751
LIQD	(5) 0.121	(3) 0.818	(4) 0.916
PLOS	(4) 0.281		(7) 0.751
FRIC	(4) 0.281	(5) 0.714	(3) 0.962
BILB		(1) 0.975	(2) 0.949
DENT		(2) 1.000	(3) 0.962
ALVR		(6) 0.282	(6) 0.806
PALT	(2) 0.513	(4) 0.787	(5) 0.860
VELR		(6) 0.282	(6) 0.806

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A test suite for hearing aid evaluation

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Abstract—A test suite has been developed for evaluating hearing aids. The tests in the suite are frequency response, number of processing bands and type of processing, input/output characteristics processing attack and release times, and broadband distortion. The test suite produces a more complete evaluation of a hearing aid than any previous set of tests, and is suitable for the automatic evaluation of a hearing aid containing unknown processing. The test procedures are described, and sample test results are presented for simulated linear and two-channel compression hearing aids.

Key words: *amplifier, automatic gain control, automatic signal processing, frequency response, hearing aid, sound pressure level, test suite.*

INTRODUCTION

A test suite has been developed for the evaluation of hearing aids and related devices. The test suite is designed to measure the characteristics of existing hearing aids and to accommodate new processing technology as it develops. The test procedures identify the signal-processing parameters of an instrument and measure overall performance.

An effective test procedure must work with the many types of signal processing available in hearing aids. The most common instruments are single-channel, containing linear processing that shapes the frequency response and provides gain. Automatic gain control (AGC) is available in some hearing aids, typically as a circuit that monitors the input signal level and provides linear gain for an input below the selected threshold and a fixed limiting output level for a signal above the threshold. Newer signal-processing circuits, often termed automatic signal processing (ASP), can adjust the frequency response of a hearing aid in response to the spectrum and level of the incoming signal (14). More complicated instruments, having two or more signal-processing channels that may contain independent compression circuits with adjustable compression thresholds and compression ratios, are being introduced and are expected to become prevalent over the next several years.

Test procedures, however, often lag behind the advances in hearing aid technology. The manufacturers' specifications for a hearing aid, for example, are commonly determined in accordance with the hearing aid test procedures given in American National Standards Institute ANSI S3.22 (1). The ANSI standards are not meant to be all-inclusive; rather, they are primarily quality control standards for the manufacture of the instruments and were originally developed for single-channel hearing aids containing linear or compression processing. More complicated processing, such as ASP or two-channel compression, cannot be adequately tested using the ANSI S3.22 procedures. Thus new tests are needed for the evaluation of the newest generation of hearing aids.

The characterization of a hearing aid requires a series of tests, since no one test can give a complete description of how the processing behaves. In developing the test suite, existing tests have been used where relevant and new tests have been derived where existing tests are inadequate. A

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total of five tests make up the test suite:

1. Frequency response as a function of input level
2. Type of processing and number of bands
3. Input/output characteristics for each band
4. Attack and release times for each band
5. Broadband distortion as a function of input level.

Shaped broadband noise signals are used for the frequency-response and distortion tests, and noise combined with a swept bias tone is used for determining the type of processing. Sinusoids stepped in amplitude are used for the input/output and attack/release measurements, with the test-frequency selection based on the results of the determination of the number and spacing of the processing bands.

The test suite has been implemented on a general-purpose personal computer as a set of programs that generate the test signals and analyze the hearing aid response. A complete hearing aid test system would require the addition of a digital-to-analog (D/A) converter and amplifier to apply the test signal to the hearing aid, an acoustic test box, and an amplifier and analog-to-digital (A/D) converter to acquire the hearing aid response in a form suitable for digital analysis. Such a test system is being built as part of a related project. In the meantime, the test suite has been evaluated using a digital simulation of a two-channel hearing aid having independent compression in each channel and including amplitude clipping in the amplifier.

The purpose of this paper is to describe the test suite and to illustrate the analysis results for typical hearing aids. The paper starts with a description of each of the five tests, including the objective of the test, the test signal and method of generation, and the signal-processing procedure for the analysis. The parameters of the simulated linear and two-channel compression hearing aids used to illustrate the tests are then presented. This is followed by a presentation of the test results, showing the test performance and indicating how hearing aid processing and adjustments can affect the analysis.

TEST PROCEDURES

The hearing aid test procedures are designed to form a practical computer-based measurement system. The concern with a practical implementation is reflected in the assumed accuracy of the D/A and A/D converters. Since the greatest level of accuracy readily available in computer-based data-acquisition systems is 16 bits, all of the procedures have been designed to work with 16-bit A/D and D/A converters. The simulation results presented here include 16-bit quantization of the test signals and hearing aid output.

The test signals span a range from 60 through 90 dB sound pressure level (SPL), and these signals have to fit within the dynamic range limitations of the 16-bit D/A converter. The assumed maximum output of the system without saturating the D/A converter is a sinusoid having a level of 102 dB SPL, which gives enough dynamic range for the output of a 90 dB SPL noise signal without clipping, given a crest factor (peak-to-average ratio) of approximately 12 dB. The dynamic range allowance for the most intense signals means that lower-level test signals will show some quantization distortion, although this will typically be below the noise level of the hearing aid under test.

Since the maximum power output (MPO) of different hearing aids can cover quite a wide dynamic range, the measurement system will need an attenuator to match the output of the hearing aid under test to the available dynamic range of the 16-bit A/D converter in the measurement system. The amount of attenuation needed to match the maximum hearing aid output signal to the full-scale input level of the A/D converter can be determined at the beginning of the set of tests and stored to give a scale factor for adjusting the response curves.

A related concern is the sampling rate used for the signal generation and analysis. The bandwidth of most hearing aids is limited by the receiver, which typically rolls off rapidly above 5 or 6 kHz (12). Thus, a sampling rate of 20 kHz was chosen to provide an adequate frequency range for a typical instrument; a higher sampling rate would require more processing time, but would not yield much additional information. Hearing aids with a bandwidth exceeding approximately 8 kHz, however, would require a higher sampling rate.

In order for the system to be practical, the processing time must also be kept within reasonable limits. In terms of the algorithms, this has resulted in the selection of single-channel analysis, based on the power spectrum of the hearing aid output, as opposed to two-channel analysis based on the cross-correlation of the output signal with the input signal. As compared to the two-channel analysis, the single-channel analysis requires half the amount of computations for a given number of data samples. A further consequence is that short test sequences have been used in order to minimize the processing time. The objective is that the complete test suite can be run in less than 20 minutes.

Frequency response

The frequency-response measurements use the shaped

Gaussian noise signal described by Burnett, *et al.* (5). The stimulus consists of white Gaussian noise that has been band-limited to the range 200 Hz to 5000 Hz, and then shaped with a one-pole low-pass filter at 900 Hz to give an approximate match to the long-term spectrum of speech. This signal is intended to test the hearing aid under conditions similar to excitation by speech. While it is obvious that the envelope fluctuations of speech are not reproduced, this test signal does come closer to a speech-like stimulus than the single swept sinusoid specified in the ANSI S3.22 standard (1).

A second reason for the use of shaped noise is to avoid the phenomenon of "blooming" that is often observed when an AGC instrument is tested with a swept sinusoid. The problem, as pointed out by Preves, *et al.* (19), is that a swept sinusoid at a constant amplitude may activate the AGC circuitry over only part of the frequency range. As the input signal level is increased, the behavior of the instrument may appear to be linear at low frequencies, where the signal level is below the AGC threshold, and compressive at high frequencies, where the AGC threshold is lower due to the frequency selectivity built into a typical AGC control circuit. Thus the frequency response appears to emphasize low frequencies as the input level is increased. The use of a broadband stimulus avoids this measurement artifact, since all frequencies are present simultaneously.

The test signal is generated from a digital pseudorandom noise (PN) generator. This gives a set of uniform random deviates between 0 and 1, which are then transformed into samples from a normal (Gaussian) distribution having zero mean and a specified output power. The white Gaussian noise is then convolved with the impulse response of the shaping filter. The filter is a digital version of the analog filter used by Burnett, *et al.* (5), and is derived in **Appendix A**. The frequency response of the digital filter is shown in **Figure 1**. The test signal is used at 60 through 90 dB SL input intensities in steps of 10 dB.

The hearing aid output is analyzed using the method of modified periodigrams (21). A total of 40,960 samples (2.05 s) are taken of the hearing aid response to the shaped noise. The first 4,096 samples (205 ms) are discarded, since these may contain transients, and the remaining samples are processed in blocks of 1,024 samples (51.2 ms), using a Hanning window and 50 percent overlap. The Hanning window gives more accurate results in regions of low relative spectral energy than other commonly used window shapes (e.g., Hamming) when used for this application. The resultant estimated power spectrum is then smoothed

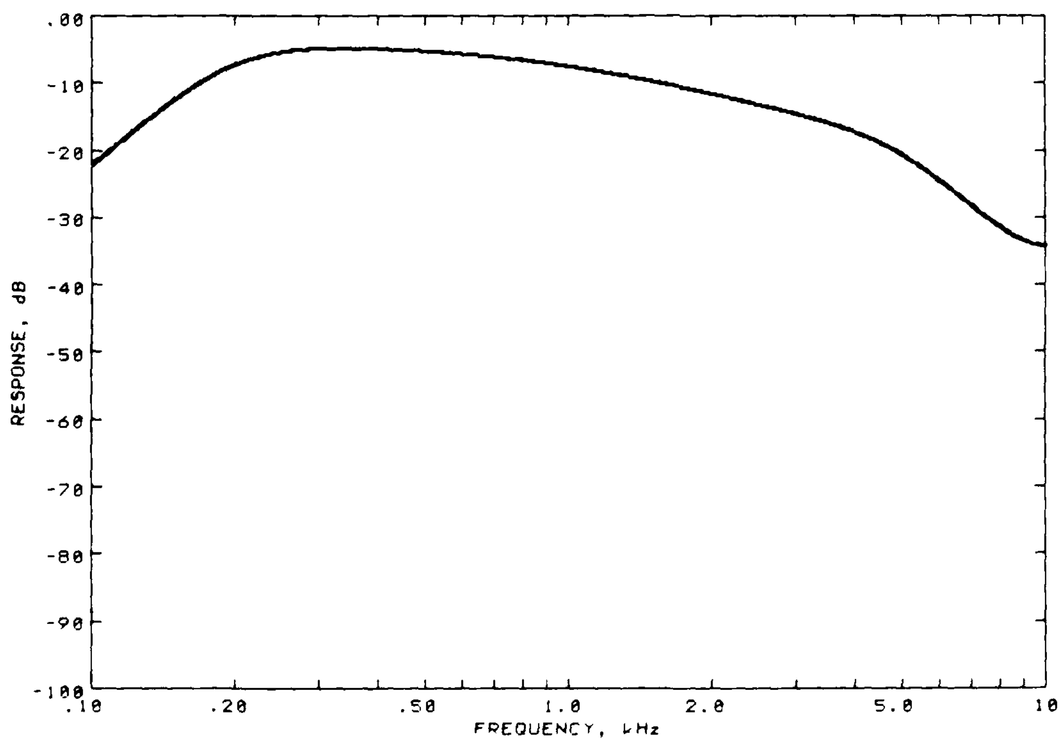

Figure 1.
Power spectrum of the digital shaped noise test signal.

by combining the frequency points into overlapping one-third octave bands, which reduces the variance in the measurement, although it will broaden sharp spectral peaks in the hearing aid frequency response. The power spectrum is divided by the spectrum of the noise-shaping filter, also smoothed by one-third octave bands, to give the gain of the hearing aid as a function of frequency. The level of the input signal in dB is then added to the gain in dB of the hearing aid to get a family of offset gain curves.

Processing type

The type of processing in the hearing aid is determined by observing how a bias tone modifies the frequency response of the instrument. To this end, the response of the instrument is measured for an excitation consisting of the 60 dB SPL-shaped noise combined with an 80 dB SPL swept sinusoid. The response of the instrument to the noise alone and to the swept tone alone are also measured and stored. The purpose of the swept tone is to bias the nonlinear processing that may be present in the hearing aid; the energy in the swept tone is detected by the control circuitry which then changes the hearing aid response. The shaped noise signal is used to determine the frequency response of the hearing aid while these changes take place.

As the swept bias tone moves through different frequency regions, it will change the gain and/or frequency response of a hearing aid containing nonlinear processing such as AGC, ASP, or amplifier saturation. Most commercially available single-channel AGC hearing aids, for example, have a bandpass filter tuned to the region around 2 kHz in front of the compressor control circuit and a threshold above 65 dB SPL in the most sensitive frequency region. An input at 80 dB SPL will be processed linearly if it is far enough removed from 2 kHz, but will cause a reduction in gain if it is near 2 kHz. The swept tone thus reduces the gain while it is present in the frequency region or regions that control the nonlinear processing. The gain will be unchanged from the linear value, however, when the swept tone is either too low or too high in frequency to be detected by the control circuitry.

To determine the changes in the frequency response caused by the nonlinear processing, the hearing aid response to the swept tone alone is subtracted from the response to the swept tone plus noise. This leaves the shaped-noise output as modified by the presence of the sweep. The spectrum of this noise is then compared to the spectrum of the hearing aid response to the shaped noise alone; the difference in the spectra indicates the degree of nonlinear processing in the hearing aid caused by the presence of the bias tone. The changes in the spectral difference as a function of the sweep frequency indicate the frequency regions controlling the nonlinear processing. A linear hearing aid will result in no recorded gain changes at any frequency of the swept tone; different realizations of equivalent linear processing, such as single-channel or multi-channel frequency shaping, will thus be indistinguishable by this procedure.

The changes in gain and frequency response are determined in 12 third-octave bands spaced from 315 to 4000 Hz at the ANSI center frequencies (2). The swept tone spends a time equivalent to 4,096 samples (205 ms) moving across each third-octave band, so the instantaneous sweep frequency has an exponential dependence on time, since the width of the bands increases with increasing center frequency. The sweep starts one third-octave below 315 Hz, with the first 4,096 samples (205 ms) discarded because of the possibility of transients, and then moves through the 12 desired bands. A total of 53,248 samples (2.66 s) are therefore analyzed, of which the last 49,152 (2.46 s) are used in determining the type of processing present in the hearing aid. The equations describing the sinusoidal sweep are given in **Appendix B**; the frequency extent of the swept tone can obviously be modified if needed.

A block diagram of the signal processing system is shown in **Figure 2**. As described above, three separate signals are sent through the hearing aid: the noise alone at 60 dB SPL, the swept tone alone at 80 dB SPL, and the shaped noise and swept tone combined. The hearing aid response to each of these signals is sampled and stored in the computer. An adaptive Widrow (22,23) noise cancellation procedure, modified for multiple reference signals (9) and adaptive estimation of the signal energy (11), is then used to estimate the changes in the system frequency response caused by the swept bias tone. The adaptive processing is described in more detail in **Appendix C**.

The adaptive noise cancellation algorithm simultaneously adjusts the filters w(n) and v(n) shown in **Figure 2** to minimize the power in the output error signal. The input to filter w(n) is the sinusoidal sweep, so very little adjustment should be needed to get nearly perfect cancellation of the sinusoidal portion of the combined signal. This adaptive filter is provided, however, since there may be situations where the presence of the shaped noise modifies the response of the hearing aid to the swept tone. The input to filter v(n) is the hearing aid response to the noise alone, so the filter v(n) has to adjust the noise-alone response to cancel the noise response in the presence of the bias sweep. The process of filtering the noise-alone response to match the noise response in the presence of the swept tone means

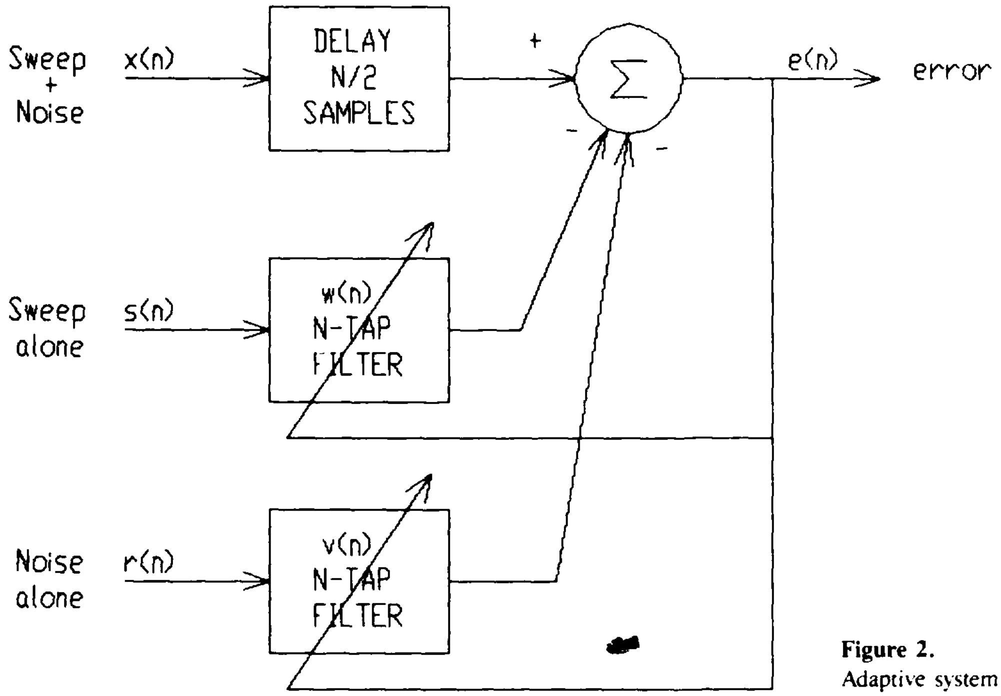

Figure 2.
Adaptive system for determining the type of processing.

that filter v(n) directly provides an estimate of the gain and frequency-response changes caused by the swept sinusoid.

The length of the adaptive filters represents an engineering compromise. A short filter will adapt more rapidly to rapid gain changes in the hearing aid, but a long filter will give better frequency resolution. For the test sequence length given above being used to test a two-channel instrument, a filter length of 15 taps has been found to be adequate. For more channels in the hearing aid, both the adaptive filter length and the duration of the test sequence should be increased so that more time is spent traversing each third-octave band.

The coefficients of filter v(n) are sampled each time the swept tone crosses the boundary of one of the 12 third-octave frequency bands and again at the end of the sweep. The frequency response is computed for each set of filter coefficients, giving a total of 13 plots. In order to simplify the presentation, the frequency response is calculated only at the frequencies corresponding to each edge of the third-octave bands. It is replaced by a blank if the reduction in gain is less than 3 dB, and by a solid character if the reduction in gain is equal to or greater than 3 dB. This leads to a square two-dimensional grid in which the gain change in each third-octave frequency region is indicated each time the sweep crosses a third-octave boundary.

A set of idealized system-identification patterns is presented in **Figure 3** for a one- or two-channel hearing aid with various types of signal processing. A perfectly linear system, as shown in the upper left-hand pattern, results in a blank pattern because the swept tone does not cause any gain change at any frequency.

The next pattern is for an ASP circuit (14). In this type of processing, the cutoff frequency of a high-pass filter moves lower in frequency as the amount of low-frequency energy decreases. The detection circuit that controls the high-pass filter tends to be most sensitive to energy around 250-300 Hz. This processing leads to a pattern in which the lower left-hand corner is blacked out on a diagonal, because raising the sweep frequency causes a reduction in the amount of energy detected by the control circuit, and as a result, the ASP cutoff frequency moves lower.

A broadband AGC circuit of the type discussed in this section results in the pattern shown in the upper right-hand corner of **Figure 3**. This pattern has a horizontal stripe blacked out since the gain at all frequencies is reduced as soon as the sweep goes above the threshold most sensitive to energy at 2 kHz. The processing is linear at low frequencies since the energy detected by the control circuit is still below the AGC threshold. Combining ASP and AGC results in a pattern where both the ASP and the AGC areas are blacked out, is shown in the lower left-hand corner.

A two-channel hearing aid having compression only in the lower frequency channel results in the next pattern. When the swept tone is in the low-frequency channel, the AGC detects the energy and causes a reduction in the gain in the channel. The gain in the high frequency is un-

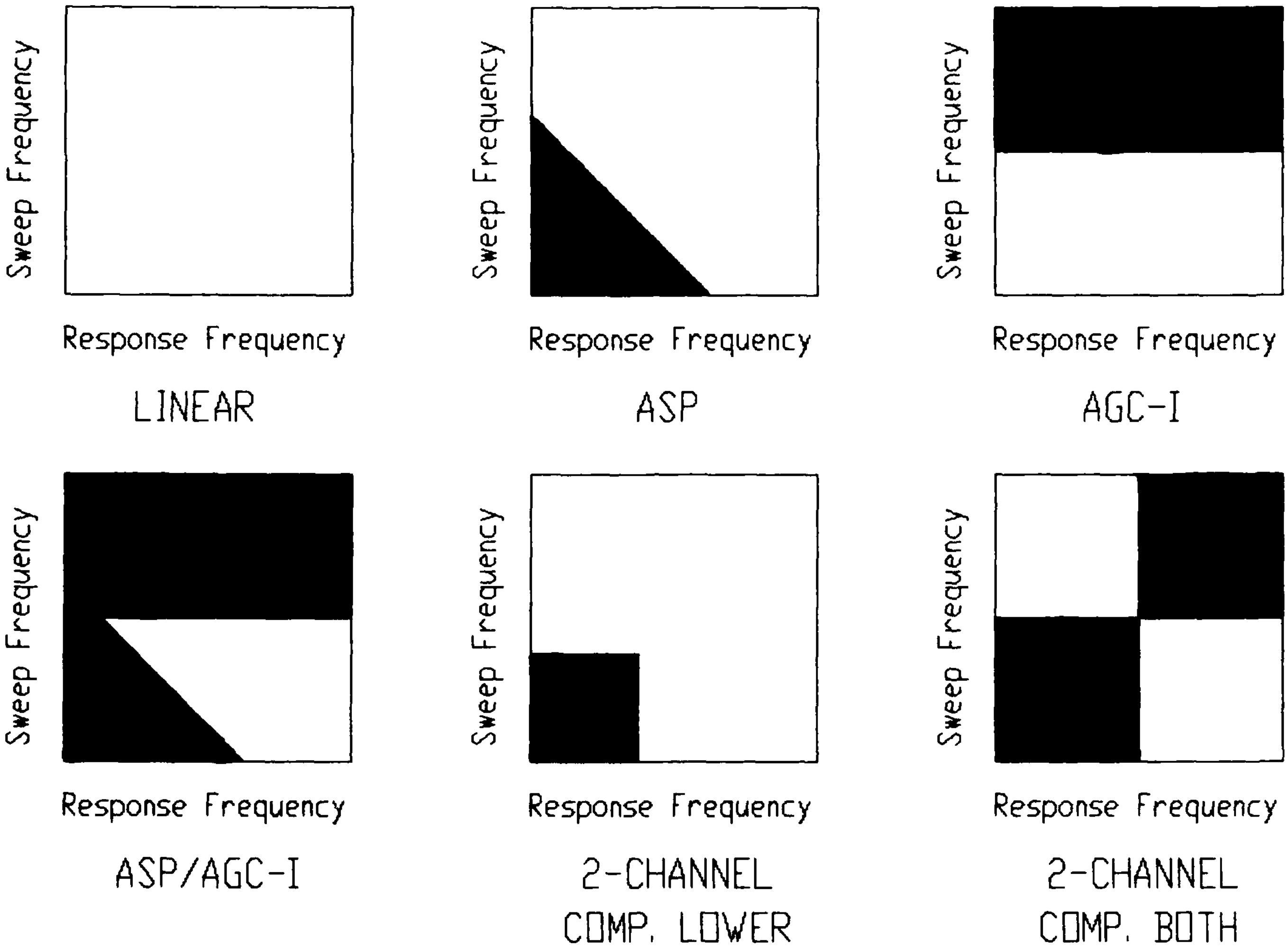

Figure 3.
Idealized system identification patterns for determining the type of processing.

changed. Thus the lower left-hand corner is blacked out because the gain at low frequencies is reduced as long as the sweep is in the lower frequency channel. The low-frequency gain returns to the linear value as soon as the sweep moves out of the low-frequency channel.

A two-channel instrument having compression in both channels results in the checker-board pattern shown in the lower right-hand corner of **Figure 3**. The gain at low frequencies is reduced while the sweep is in the low-frequency channel. The gain at high frequencies is kept at the linear value since the AGC in that channel has not been engaged. The gain at high frequencies is then reduced when the sweep moves into the high-frequency channel, and the gain at low frequencies returns to the linear value for the channel.

The test program correlates the actual system identification pattern for the hearing aid under test with each of the idealized patterns shown in **Figure 3** to return a pattern-match score. The correlation gives a value of 1 at an individual point if the actual and idealized patterns both have the same gain characteristic. It gives a value of −1 if one pattern shows no gain reduction but the other pattern does show a gain reduction at the point. The values are summed over the entire pattern and normalized by the number of points to give the test scores, which range from 1 to −1. In determining the scores for the processing options, all possible crossover frequencies are tried for each option, with the program indicating the score for the best frequency match and the crossover frequency at which it occurs. The final crossover frequency estimated by the test program is then the crossover frequency at which the highest test score was found, and the indicated processing is that which corresponds to the highest correlation score.

Input/output characteristics

In a multichannel hearing aid, the input/output characteristics are measured at or near the geometric center of each frequency band. In a single-channel instrument the measurement is made at the 2 kHz frequency used for the ANSI attack and release time measurements (1) because many single-channel compression instruments are most sensitive at this frequency. The geometric center of the low-frequency band is determined as the square root of the product of the crossover frequency and 315 Hz, and the geometric center of the high-frequency band is determined as the square root of the product of the crossover frequency and 4 kHz. The actual test frequencies are then the ANSI third-octave frequencies (2) closest to the computed band centers.

The test signal is a sinusoid of fixed frequency having segments stepped in level from 40 dB SPL to 95 dB SPL in steps of 5 dB. The duration of each segment is 250 ms (5,000 samples), with the first 100 ms an allowance for any transients to stabilize and the signal amplitude determined from the average over the last 150 ms of the segment. The test signal has 12 segments, so the total signal duration is 3 s (60,000 samples). The input level, the output level, and the compression ratio are indicated for each of the segments.

In order to reduce the amount of computer processing, the test signals are generated in advance and stored on disc. This precludes generating a test signal at each arbitrary band center frequency; instead, the precomputed signal closest to the desired center frequency is used. A set of 12 test signals, spaced at the ANSI standard (2) third-octave frequencies between 315 Hz and 4000 Hz inclusive, appears to be adequate. Note that the ANSI standard (1) test frequency of 2000 Hz is part of this set.

Attack and release times

The test procedure for measuring the attack and release times is based on the ANSI standard (1), but with provision made for analysis at the third-octave frequency closest to the geometric center of each hearing aid processing band. The test signal starts at a level of 55 dB SPL, jumps to 80 dB SPL after 200 ms (4,000 samples), and returns to 55 dB SPL at 600 ms (12,000 samples). The total signal duration is 1,000 ms (20,000 samples), but this can be lengthened for an instrument with unusually long attack or release times. The attack time is the time it takes for the hearing aid output to decay to a level 2 dB above the steady-state level after the test signal jumps to 80 dB SPL. The release time is the time it takes for the hearing aid output to reach a level 2 dB below the steady-state level after the test signal returns to the 55 dB SPL level.

Measuring the attack and release times requires extracting the envelope of the signal. Because the test signal is at a constant frequency, the simple system of **Figure 4** is appropriate. The all-pass filter is a one-zero/one-pole filter, adjusted to give 90 degrees of phase shift at the test frequency. The direct signal path thus gives the in-phase part of the signal envelope, while the output of the all-pass filter provides the quadrature component. Squaring and summing the two signals results in the magnitude-squared signal envelope used in determining the attack and release times.

Broadband distortion

The ANSI standard (1) distortion measurement recommends total harmonic distortion for input sinusoids at 500, 800, and 1600 Hz. This type of measurement does not give any indication of how distortion at high frequencies, where many hearing aids have their highest amount of gain, and therefore most easily go into amplifier saturation, will affect the reproduction of lower frequency speech sounds occurring at the same time. A procedure for measuring intermodulation distortion for a broadband stimulus, on the other hand, can provide information on the amount of

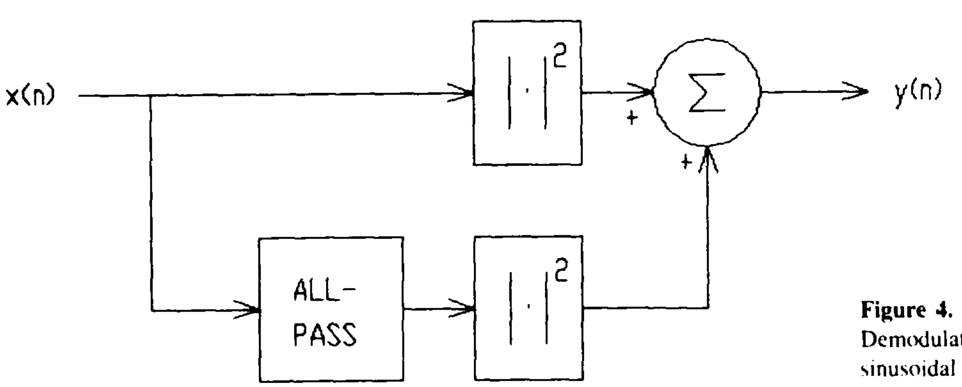

Figure 4.
Demodulation system for extracting the envelope of a sinusoidal signal.

distortion that will occur under typical listening conditions.

Noise distortion measurements have been proposed before (4,18). These procedures used a broadband stimulus convolved with an analog filter having a single notch in the transfer function. The noise distortion was computed from the amount of energy that was found in the notch. The system used in the test suite differs from these earlier proposals in that multiple notches are placed in the test signal rather than the single notch, and the test-signal generation is based on digital signal processing techniques.

The test signal is the shaped noise used for the frequency-response measurements and which is then convolved with a comb filter to create a series of interleaved peaks and valleys. The comb filter is a half-band filter designed using the Parks-McClellan design program (16) and having 0 dB pass-band gain and 62 dB stop-band attenuation, so it is theoretically possible to measure distortion values of less than 0.1 percent. Replacing the unit delay in the half-band high-pass filter with a delay of two samples results in the bandpass filter response shown in **Figure 5**, illustrating the ripple in the pass-band and stop-band of the filter. Replacing the unit delay with a delay of 32 samples results in the filter response shown in **Figure 6**, which now has a set of 16 peaks and valleys uniformly spaced in frequency from 0 to 10 kHz. The filter coefficients are given in **Appendix D**; the comb filter has a total length of 961 taps, but is still computationally efficient since only 17 of these taps are non-zero.

The distortion measurement procedure is based on determining how much energy from the peaks of the comb-filtered shaped noise signal spills over into the valleys. A total of 40,960 samples (2.05 s) of the hearing aid response to the test signal are acquired. As for the frequency-response measurements, and the first 4,096 samples (205 ms) are again discarded since they may be contaminated by transients. The power spectrum is computed using the method of modified periodigrams (21), with the data processed in blocks of 2,048 samples (102 ms) using a Hanning window and 50 percent overlap. The block length and type of window are important, since it is critical that leakage from the peaks of the spectrum not bias the estimated distortion levels in the valleys of the test signal. The distortion energy is computed as the mean-squared energy in the center of the valley, and the signal energy is computed as the mean-squared energy in the peaks to either side of the valley with the bias equal to the distortion energy subtracted out. The transition regions between the peaks and the valley are not used. The signal-to-distortion ratio (SDR) is then the square root of the ratio of the computed distortion energy to the signal energy, expressed as percent or in dB for each valley frequency.

Table 1.
Weights for combining the signal-to-distortion ratios into a single distortion index.

Band	Valley Center Frequency, Hz	Band Edges, Hz	Weight
1.	625	312.5- 937.5	0.258
2.	1250	937.5-1562.5	0.209
3.	1875	1562.5-2187.5	0.180
4.	2500	2187.5-2812.5	0.121
5.	3125	2812.5-3437.5	0.084
6.	3750	3437.5-4062.5	0.063
7.	4375	4062.5-4687.5	0.047
8.	5000	4687.5-5312.5	0.038

In addition to expressing the distortion as a function of frequency, there is a need for a single figure of merit to indicate the overall effect of the distortion on intelligibility. The figure of merit used in the broadband distortion measurement is a weighted SDR, computed using a procedure similar to that of the Articulation Index (10,15). The SDRs are computed in dB for each valley frequency. A ratio greater than 30 dB is set to 30 dB, and a ratio less than 0 dB is set to 0 dB. The ratios are then multiplied by the weighting factors given in **Table 1**, summed, and divided by 30 to get a number between 0 and 1. The weights are based on the number of the equally important bands for speech intelligibility that are included in each of the distortion bands. The weights from French and Steinberg (10) are for nonsense syllables; modifications to the weights can be made for connected discourse (17), but the change in the weighted SDR is small.

The above distortion measurement procedure was chosen over the magnitude-squared coherence function (3,7) because of bias effects that can occur in computing the coherence. The magnitude-squared coherence function is the magnitude-squared cross-spectral density of the input and output signals, normalized by the product of the spectral densities of each of the two signals. The function is computed in the frequency domain by dividing the data into segments and averaging the appropriate fast Fourier transform (FFT) products over the segments. The coherence is high if the phase differences between the output and input remain constant from segment to segment as would occur for an ideal linear system, and is low if the phase differences fluctuate as would occur for noise or uncorrelated distortion.

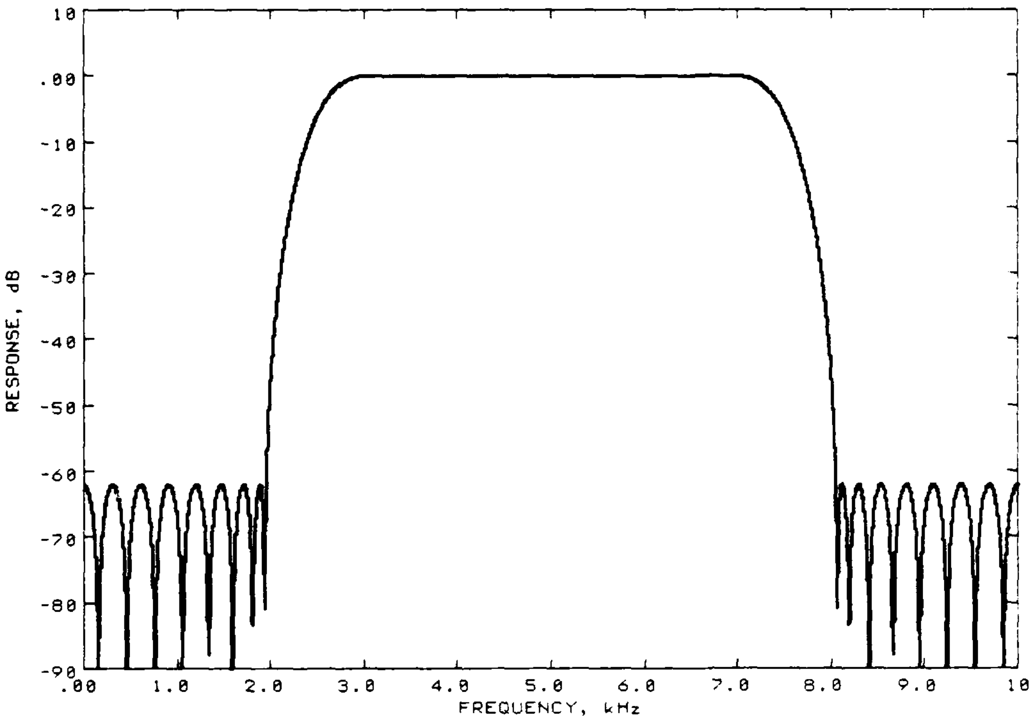

Figure 5.
Frequency response of the half-band filter having the unit delay replaced by a two-sample delay.

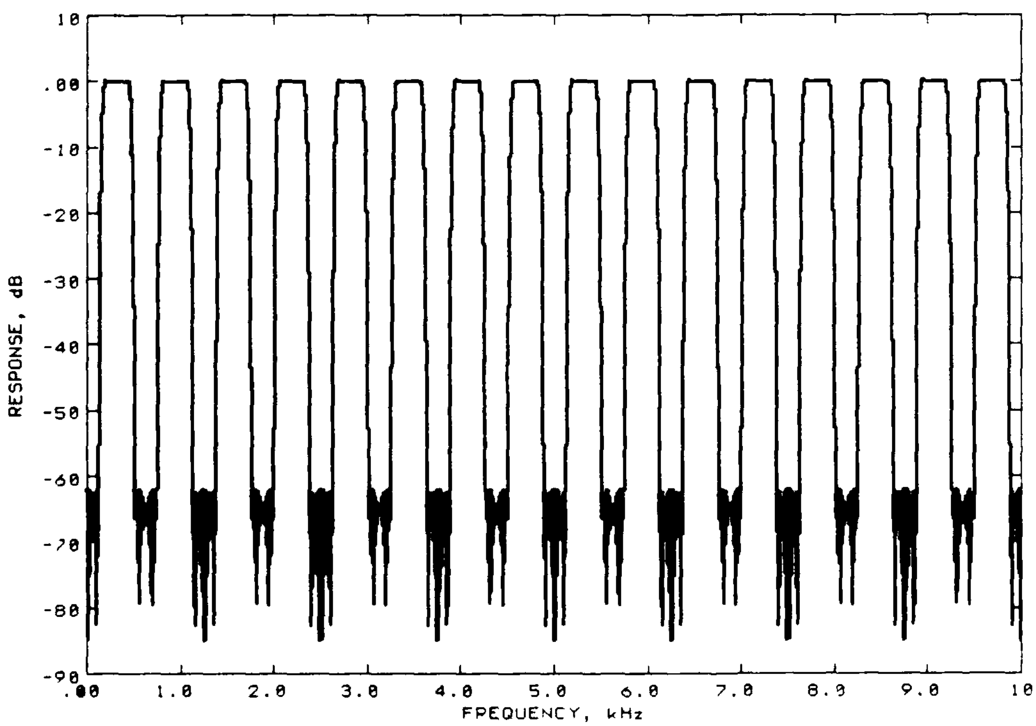

Figure 6.
Frequency response of the comb filter derived from the half-band filter by replacing the unit delay with a delay of 32 samples to give 16 combs.

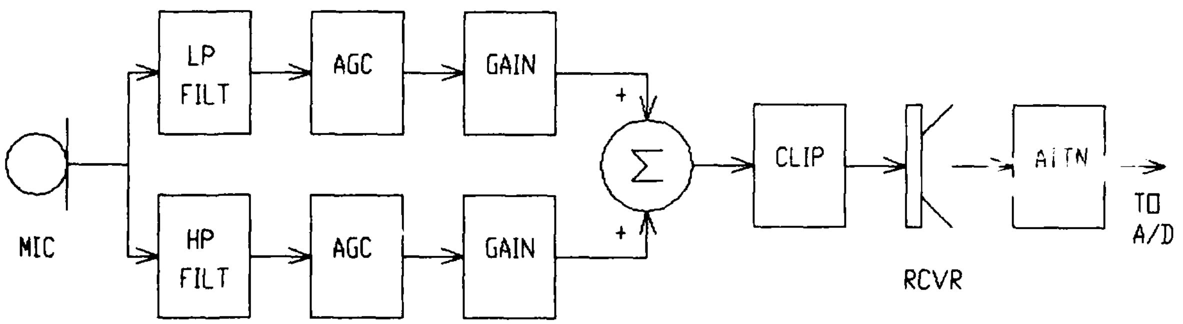

Figure 7.
Block diagram of the simulated hearing aid.

The bias in estimating the coherence function is inversely proportional to the number of data segmenpts used in its estimation (7,8). This can be seen on an intuitive basis since the coherence function for a single segment is identically unity (no phase fluctuations can be detected if only one sample of the phase difference is available), so many segments will have to be processed in order to get an accurate estimate of a low coherence function value. Using small segments in order to reduce this bias, however, leads to a different set of problems, since misalignment of the time delays between the signals being cross-correlated leads to a bias that is inversely proportional to the segment length (6,20). A related bias effect comes from the truncation of the system impulse response by the limited segment length. Examples comparing the distortion estimated using comb-filtered noise and via the magnitude-squared coherence function are presented later in this paper.

HEARING AID SIMULATIONS

A digital simulation of a hearing aid is used to illustrate the test processing; the time-domain simulation is described by Kates (13) and a brief summary is given here. A block diagram of the simulated instrument is presented in **Figure 7**. The input to the microphone is the free-field sound pressure generated by an ideal loudspeaker. The test signal is the amplified 16-bit output of the computer D/A converter. The microphone output is split into a low-pass and a high-pass channel, with independent input-referred compression (AGC-I) in each channel. The compression ratio and compression threshold can be specified for each channel. The gain in each channel is then adjusted, and the signals summed. The power amplifier is represented as a gain of 0 dB followed by symmetric output clipping. The receiver is assumed to be loaded acoustically by a 2 cm^3 coupler; all other acoustic tubing and venting is ignored, so the output is representative of an in-the-ear (ITE) hearing aid as measured in a test chamber. The coupler output passes through an attenuator prior to being sampled by the 16-bit computer A/D converter. The computations for the simulation itself are all carried out using floating-point arithmetic.

Four simulated hearing aids are used to illustrate the test procedures. Two are linear instruments, and two are compression instruments. The linear hearing aid L00 is described by the parameters given in **Table 2**. This is a type of instrument commonly fitted for flat hearing losses. The two channels in this hearing aid have equal gains, and are equivalent to a single channel with a flat frequency response. Amplifier saturation is modeled by the clipping of all samples above a peak magnitude of 85 dB SPL less the microphone and channel gains. This distortion threshold is typical of commercially available hearing aids, and the symmetrical clipping is a reasonable model of amplifier saturation (13).

The second linear hearing aid L20 is identical to the first except that the low-frequency gain is set to −20 dB. This type of frequency response is commonly fitted for sloping hearing losses. The two linear channels are equivalent to a single-channel hearing aid with frequency shaping in the microphone response or in a preamplifier stage prior to the final amplifier. Again, amplifier saturation is modeled by the clipping of all samples above a peak magnitude of 85 dB less the microphone and channel gains.

The second pair of simulated hearing aids are two-channel compression instruments. Hearing aid C00 is given by the parameters of **Table 3**. This hearing aid is typical of the newest products being introduced by the industry. The crossover frequency and the specified compression ratios and thresholds in each channel are meant for illustrative purposes only and do not necessarily represent a recommended set of parameters for an actual fitting. The

two channels have equal gains, and are therefore typical of an instrument that would be fitted for a flat loss. The final amplifier stage and receiver in the simulated compression hearing aid are assumed to be the same as in the linear instrument, and the same clipping threshold for the instrument L00 is used.

The second compression hearing aid C20 is identical to the first except that the low-frequency gain is set to −20 dB. This simulated instrument has a linear (low-level) frequency response typical to that fitted for a sloping hearing loss. All other parameters are the same as for hearing aid C00.

EXAMPLES AND DISCUSSION

Frequency response

The frequency-response test produces a family of response curves for shaped-noise input signals ranging from 60 dB SPL to 90 dB SPL in steps of 10 dB. The sets of offset gain curves for the four simulated hearing aids are presented in **Figure 8**. The linear behavior of instrument L00 is shown by the uniform 10 dB separation between the lower three curves in **Figure 8a**, and the saturation of the amplifier becomes apparent for the 90 dB SPL input level since the last increment of 10 dB in the input results in about a 3 dB increase in the output at all frequencies. The linear hearing aid L20 shown in **Figure 8b** also exhibits the effects of saturation at the 90 dB SPL input level, but the effects are slightly less pronounced since the −20 dB gain in the low-frequency channel results in less power reaching the amplifier for the same input stimulus level.

Table 2.
Parameters for the simulated linear hearing aid L00.

Microphone:	Knowles EA-1842
Receiver:	Knowles ED-1913
Amplifier type:	Current source
Amplifier clipping level:	85 dB SPL
Crossover frequency:	2 kHz
Low-frequency channel	
Gain:	0 dB
Compression ratio:	1:1
Compression threshold:	Does not apply
Attack time:	Does not apply
Release time:	Does not apply
High-frequency channel	
Gain:	0 dB
Compression ratio:	1:1
Compression threshold:	Does not apply
Attack time:	Does not apply
Release time:	Does not apply

Table 3.
Parameters for the simulated compression hearing aid C00.

Microphone:	Knowles EA-1842
Receiver:	Knowles ED-1913
Amplifier type:	Current source
Amplifier clipping level:	85 dB SPL
Crossover frequency:	2 kHz
Low-frequency channel	
Gain:	0 dB
Compression ratio:	4:1
Compression threshold:	75 dB SPL
Attack time:	1 msec
Release time:	50 msec
High-frequency channel	
Gain:	0 dB
Compression ratio:	2:1
Compression threshold:	65 dB SPL
Attack time:	1 msec
Release time:	50 msec

The families of frequency-response curves for the simulated compression instruments C00 and C20 are shown in **Figures 8c** and **8d**, respectively. At low frequencies, one can see the linear increase in gain for the input signal below the compression threshold of 75 dB SPL, followed by the closer spacing of the output curves as the input increases in level due to the 4:1 compression that is engaged for signals above threshold. At frequencies above the 2 kHz crossover, the spacing of the curves reflects the lower threshold and 2:1 compression ratio chosen for the high-frequency channel. The combination of compression thresholds and compression ratios chosen for the two channels results in the response curve for the 90 dB SPL stimulus level being parallel to the response curve for the 60 dB SPL stimulus level for both simulated compression hearing aids. The measurements thus give an accurate indication of the frequency-response behavior of the hearing aid.

Processing type

The test to determine the type of processing and the number of bands produces a system identification pattern giving the response of the instrument under test as function of the sweep frequency. The patterns produced by the test program are quantized in level, with a gain reduction of less than 3 dB indicated by a dot, and a gain reduction of greater than 3 dB indicated by the symbol "#." The resultant pattern is a 13 × 13 grid, since the adaptive filter coefficients are sampled each time the sweep frequency crosses the edge of one of the third-octave bands and the

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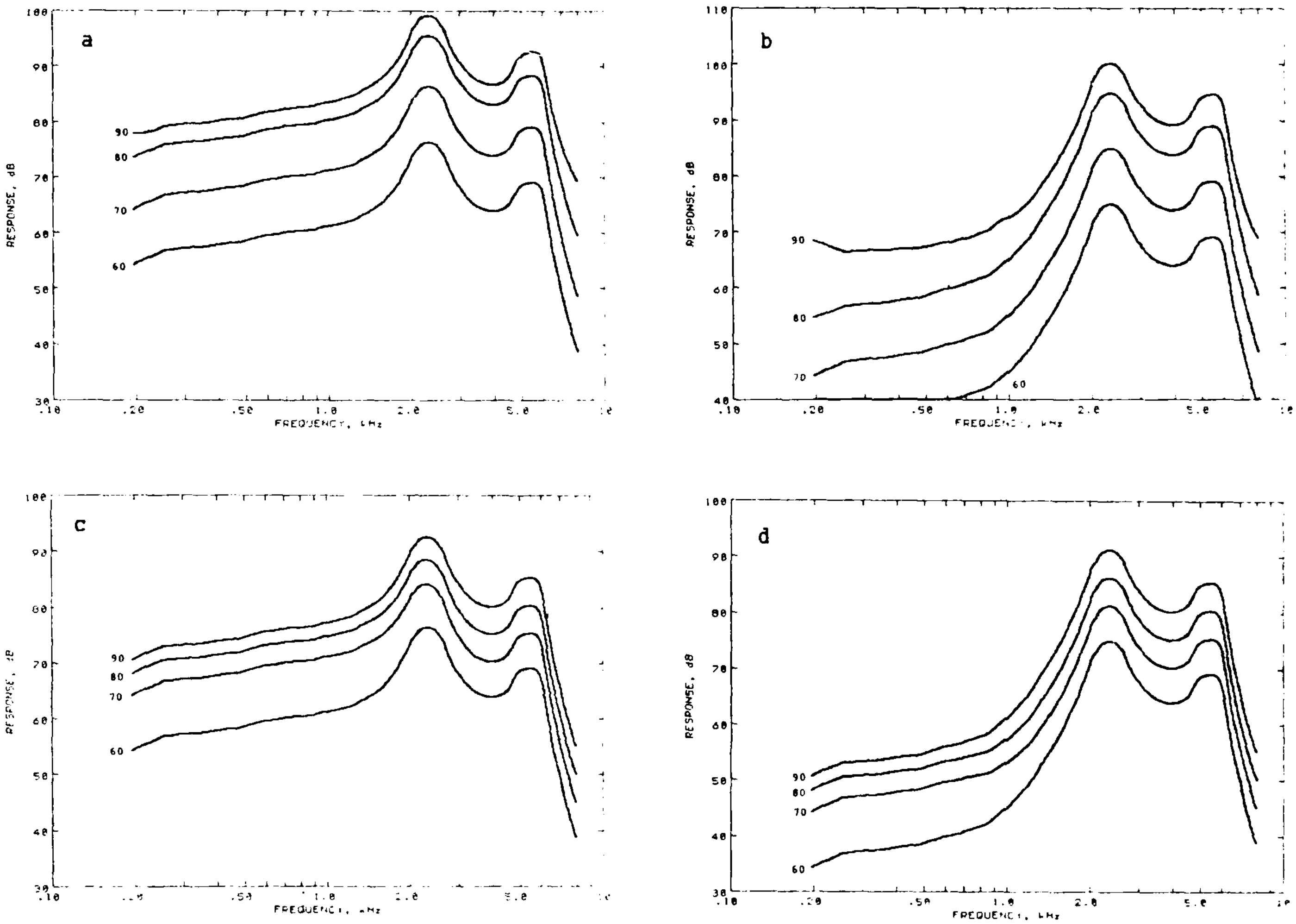

Figure 8.
Set of offset gain curves for input levels from 60 dB SPL to 90 dB SPL in steps of 10 dB for: a) the simulated linear hearing aid L00; b) the simulated linear hearing aid L20; c) the simulated compression aid C00; and, d) the simulated compression aid C20.

filter frequency-response values are calculated for frequencies that are also at the edge of each of the 12 third-octave bands. The length of the adaptive filters used for the processing was set to 15 taps.

The system identification pattern for the simulated linear hearing aid L00 is shown in **Figure 9a**. The linear hearing aid processing results in essentially no gain reduction due to the swept bias tone. The one exception to this statement occurs at the top of the pattern, where the bias tone is amplified slightly by the 5 kHz resonance peak of the microphone response and this causes some clipping in the amplifier. This clipping results in the apparent gain reduction at all frequencies shown in **Figure 9**. The pattern match scores show a maximum for AGC processing, even though this is nominally a linear hearing aid, and indicate that the gain reduction occurs when the swept tone goes above 3150 Hz.

The system identification pattern for the linear hearing aid L20 is shown in **Figure 9b**. The pattern is similar to that for the flat linear-response instrument, except that the amplifier saturation only reduces the gain in the vicinity of 2 kHz. The highest pattern match score is now for linear processing. Thus, changing the frequency response can affect the classification of a nominal linear instrument by changing the saturation effects.

The processing built into the simulated compression hearing aid C00 is analyzed in **Figure 9c**. One can see the extent of the low-frequency channel, because there is a gain reduction everywhere in the channel for as long as the sweep is within it. As the sweep moves through the crossover region of the instrument, there is a gain reduction in both channels, and finally the high-frequency channel alone is affected as the sweep moves higher in frequency. Two-band compression has the highest processing score,

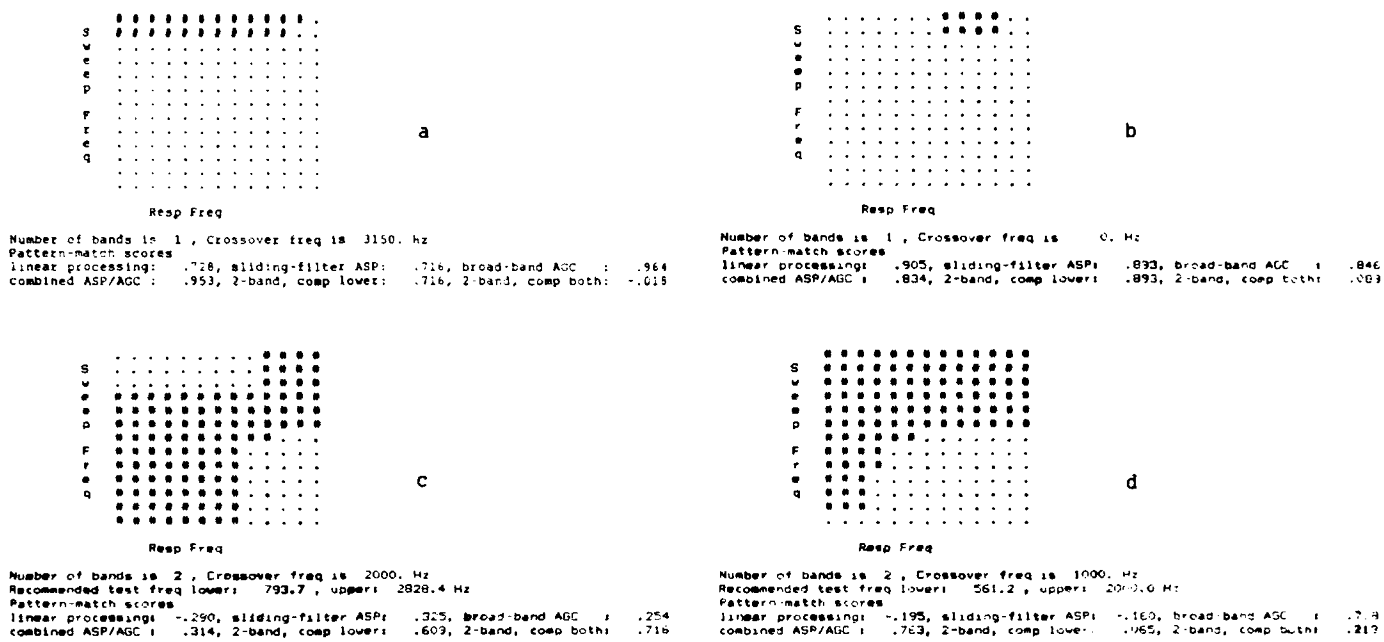

Figure 9.
System identification pattern and scores for: a) the simulated linear hearing aid L00; b) the simulated linear hearing aid L20; c) the simulated compression aid C00; and, d) the simulated compression aid C20.

and the program has correctly identified the 2 kHz crossover frequency as well as the number of processing bands.

The pattern for the simulated compression hearing aid C20 is shown in **Figure 9d**. The pattern shows that when the sweep frequency is within the low-frequency channel, there is a gain reduction. The reduced gain, however, only appears in the lower half of the low-frequency channel. When the sweep is within the high-frequency channel, there appears to be a gain reduction at all frequencies rather than just in the high-frequency channel. The pattern matching scores reflect this behavior, since the broadband AGC and ASP/AGC both give higher scores than two-channel compression. The test pattern is a consequence of the high-frequency channel being given 20 dB more gain in the simulated instrument than the low-frequency channel. Because of this gain differential, the high-frequency channel dominates the frequency response of the instrument and the gain changes in the low-frequency channel have only a limited effect. Thus a two-channel compression instrument behaves more and more like a single-channel AGC instrument as the gain in one channel is increased relative to the gain in the other.

Input/output characteristics

The input/output characteristics for a linear hearing aid are determined at the ANSI (1) test frequency of 2 kHz. The characteristics for a two-channel hearing aid are determined at the geometric center frequencies of each of the frequency bands found by the processing test. The crossover frequency was found to be 2 kHz for the two-channel compression instrument C00. The closest third-octave test frequency for the low-frequency band is 800 Hz, and for the high-frequency band, 3150 Hz.

The input/output characteristics for the linear hearing aid L00 are given in **Table 4**. The linear operation is clearly indicated by the compression ratio of 1.0 for inputs up to 85 dB SPL. At higher stimulus intensities the amplifier clipping becomes increasingly more prominent, and the output saturation is reflected in the indicated high compression ratios. The behavior of the linear instrument L20 is similar, as shown by the data of **Table 5**, although the clipping is not quite as pronounced due to the −20 dB low-frequency gain of this simulated hearing aid.

The input/output characteristics for the compression hearing aid C00 are presented in **Table 6** for the test signals at 800 Hz and 3150 Hz. With equal 0 dB gains for the two channels of the compression instrument, the low-frequency channel measurement comes close to showing the 4:1 compression ratio starting at a threshold of 75 dB SPL specified for this simulation. The high-frequency channel shows some minor influence from the low-frequency channel as its compression is engaged, but the measured compression

Table 4.
Input/output measurements for the simulated linear hearing aid L00.

Input dB SPL	Freq = 2000 Hz Output dB SPL	Comp Ratio
40	52.01	—
45	56.96	1.01
50	61.93	1.01
55	66.97	0.99
60	71.99	0.99
65	77.00	1.00
70	82.00	1.00
75	86.99	1.00
80	91.99	1.00
85	94.11	2.36
90	94.66	9.15
95	94.93	18.04

Table 5.
Input/output measurements for the simulated linear hearing aid L20.

Input dB SPL	Freq = 2000 Hz Output dB SPL	Comp Ratio
40	48.93	—
45	54.00	0.99
50	58.94	1.01
55	64.03	0.98
60	69.03	1.00
65	74.03	1.00
70	79.03	1.00
75	84.03	1.00
80	89.02	1.00
85	93.52	1.11
90	94.32	6.20
95	94.90	8.63

ratio is still reasonably close to the specified value of 2:1.

The input/output characteristics at the same pair of test frequencies for the simulated two-channel compression hearing aid C20 are given in **Table 7**. The compression ratio of 1.40 for the 45 dB SPL input signal level is an artifact caused by the A/D quantization of the low hearing aid output level since the low-frequency gain is −20 dB. The compression ratio then becomes nearly unity over the linear region of the instrument, and becomes greater than one for input signal levels above the compression threshold. The compression ratio is not the 4:1 set in the parameters for the compression hearing aid, but rather a lower value because of the strong influence of the high-frequency channel due to its 20 dB greater gain. The input/output characteristics for the high-frequency channel tested at 3150 Hz very clearly show the compression threshold and 2:1 compression ratio set for the simulation because the reduced gain in the low-frequency channel minimizes its influence on the measured characteristics of the high-frequency channel.

The clearest indication of the behavior of an individual compression channel can be found when the other channel is turned off. Otherwise the measured input/output characteristics will tend to reflect the dominant frequency band no matter which channel is being measured. As the gain in one channel is increased relative to the other, the hearing aid behaves more and more like a single-channel instrument. Thus the relative gains in the different frequency channels can be an important factor in the measured compression ratios and thresholds.

Attack and release times

The measured attack and release times depend on both the system under test and the measurement procedure. The presence of the all-pass filter shown in the envelope detection system of **Figure 4** will affect the estimated attack and release times due to the transient behavior of the analysis filter itself. Additional transient effects come from the inherent linear behavior of the hearing aid, particularly from the receiver response. The degree of overshoot for the computed envelope of the output of the linear hearing aid L00, excited by the stepped 2 kHz excitation, is shown in **Figure 10a**. The behavior of linear instrument L20 in **Figure 10b** is quite similar. There is less than 2 dB of overshoot on the attack portion of the signal, and less than 2 dB of undershoot on the release portion. This is below the criterion for the ANSI S3.22 test procedure, so the reported attack and release times are both zero. Neither the envelope computation nor the behavior of the linear elements of the hearing aid will significantly bias the calculation of the attack and release times.

The envelopes are computed at both test frequencies

Table 6.
Input/output measurements for the simulated compression hearing aid C00.

	Freq = 800 Hz		Freq = 3150 Hz	
Input dB SPL	Output dB SPL	Comp Ratio	Output dB SPL	Comp Ratio
40	40.33	—	45.98	—
45	45.20	1.03	50.96	1.01
50	50.21	1.00	55.95	1.00
55	55.21	1.00	60.96	1.00
60	60.22	1.00	65.67	1.06
65	65.21	1.00	68.33	1.88
70	70.22	1.00	71.10	1.80
75	73.70	1.43	74.05	1.70
80	74.98	3.91	77.25	1.56
85	76.30	3.80	80.00	1.82
90	77.64	3.72	82.07	2.41
95	78.96	3.80	84.22	2.33

Table 7.
Input/output measurements for the simulated compression hearing aid C20.

	Freq = 800 Hz		Freq = 3150 Hz	
Input dB SPL	Output dB SPL	Comp Ratio	Output dB SPL	Comp Ratio
40	23.27	—	45.77	—
45	26.85	1.40	50.77	1.00
50	31.75	1.02	55.75	1.00
55	36.43	1.07	60.76	1.00
60	41.52	0.98	65.45	1.07
65	46.54	1.00	67.95	2.00
70	51.49	1.01	70.46	2.00
75	55.41	1.28	72.96	1.99
80	58.28	1.74	75.47	1.99
85	61.94	1.37	77.98	2.00
90	65.19	1.54	80.47	2.01
95	67.49	2.17	82.97	2.00

for the two simulated two-channel compression hearing aids. The envelope for the 800 Hz test frequency, which excites primarily the low-frequency channel, is shown in **Figure 10c** for instrument C00, and in **Figure 10e** for instrument C20. The measured attack times are 2.6 ms for C00 and 1.4 ms for C20. These values are reasonably close to the specified time of 1 ms, but the measured release times of 10.3 and 12.7 ms for C00 and C20, respectively, are much shorter than the specified release time of 50 ms.

The difference between the estimated and specified release times is a result of the compression threshold of 75 dB SPL set in the low-frequency channel. This threshold is close to the upper signal level of 80 dB SPL. After the test signal returns to the 55 dB SPL level, the hearing aid spends only a short period of time in compression before the detection circuit indicates that the gain should return to the linear value used for inputs below the compression threshold. Thus the measured release time is a function of the compression threshold: the higher the threshold, the shorter the apparent release time, all else being held constant.

The test signal at 3150 Hz excites primarily the high-frequency channel. The envelopes for simulated compression instruments C00 and C20 are shown in **Figures 10d** and **10f**, respectively. The measured attack times are 1.65 ms for instrument C00 and 1.75 ms for C20, and the measured release times are 53.8 ms for C00 and 50.9 ms for C20, respectively. These measured values are close to the specified attack time of 1 ms and release time of 50 ms. The longer release times are due to the lower compression threshold in the high-frequency channel. However, the effect of the compression threshold on the attack times appears to be much smaller.

Broadband distortion

The broadband distortion measurement determines the amount of energy in the valleys of the comb-filtered shaped noise signal relative to the energy in the two adjacent peaks. The detection threshold depends on the quantization noise in the computer D/A and A/D converters as well as the side-lobe levels of the comb filter and the round-off error and leakage in the discreet Fourier transform used for the spectral analysis. The internal noise floor of a typical hearing aid, not included in the simulation, will also affect the measured distortion levels, because the measurement procedure cannot distinguish between distortion and noise. The proper levels for the test signal represent a trade-off between the desire on the one hand for a high signal level to minimize the relative importance of the internal and measurement noise, and the desire on the other hand for a low signal level to insure at least one set of measurements in the linear operating range of the hearing aid.

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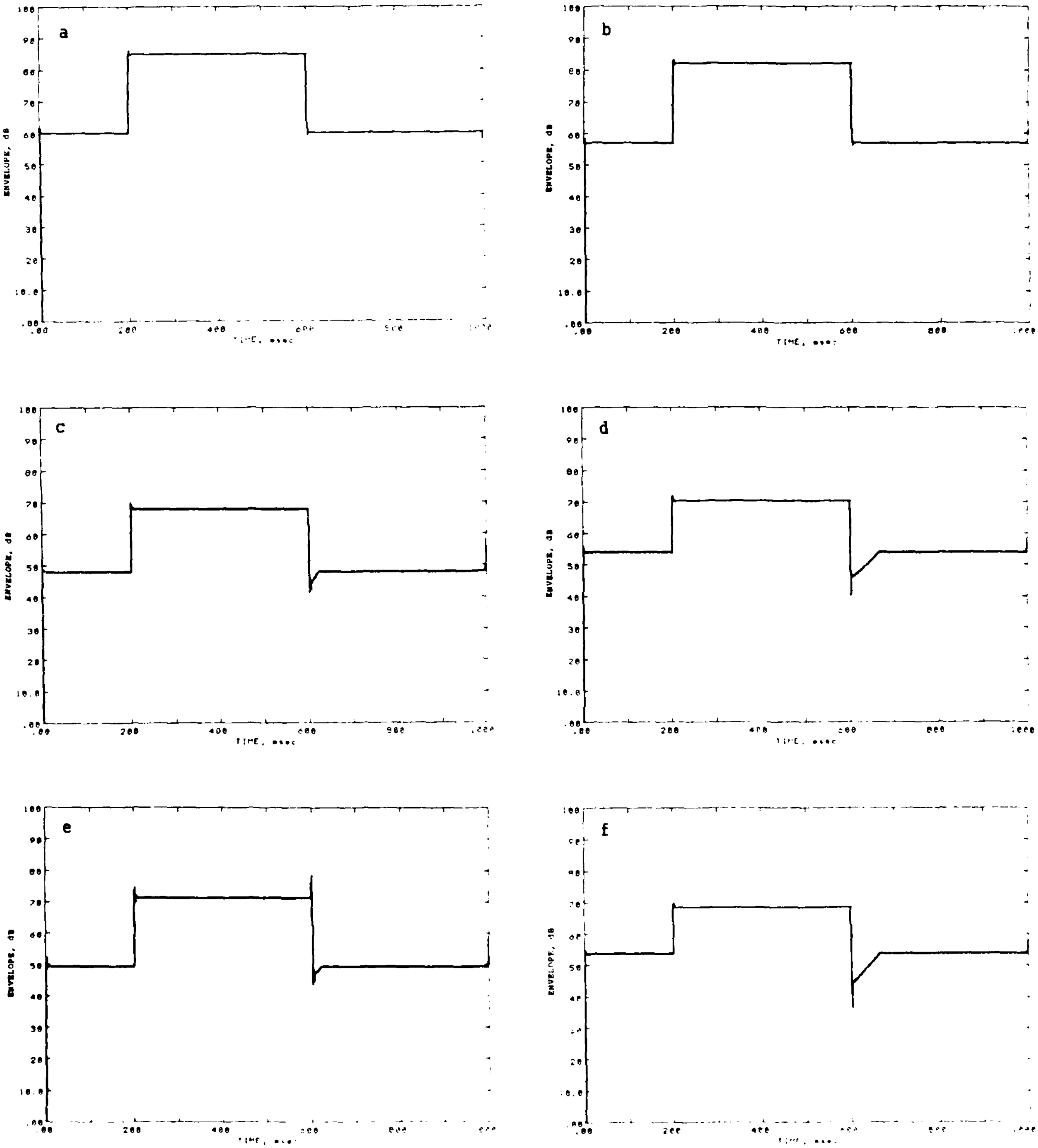

Figure 10.
Signal envelope from the attack and release time measurements of: a) the simulated linear hearing aid L00 measured at 2000 Hz; b) the simulated linear hearing aid L20 measured at 2000 Hz; c) the simulated compression aid C00 measured at 800 Hz; d) the simulated compression aid C00 measured at 3150 Hz; e) the simulated compression aid C20 measured at 800 Hz; and, f) the simulated compression aid C20 measured at 3150 Hz.

A set of broadband distortion measurements for the simulated linear hearing aid L00 is presented in **Table 8**, and the corresponding data for the linear instrument L20 is presented in **Table 9**. The percent distortion in the valleys from 625 through 5000 Hz are given for input signal levels from 60 through 90 dB SPL, along with the weighted SDR index. The lowest distortion values are for the input signal at 70 dB SPL, not at 60 dB SPL; both signal levels are well below the amplifier clipping threshold, but the lower input level is affected more by the quantization noise of the digital sampling process. The response of linear instruments L00 and L20 to the 70 dB SPL input signal is shown in **Figure 11a** and **Figure 11b**, respectively, where the energy in the valleys falls below the frequency-response axis. The bulk of the measured values for the 70 dB SPL input signal show that the procedure is capable of measuring distortion at the level of 0.1 percent when the signal level is carefully adjusted.

Increasing the level of the test signal above 70 dB SPL results in increased distortion. The data for the 80 dB SPL test signal presented in **Table 8** and **Table 9** shows a substantial increase in the measured distortion when compared with the lower signal levels. Since the distortion levels are still relatively low, however, the weighted SDR index only shows a modest decrease to the value of 0.821 for instrument L00, and 0.951 for instrument L20.

Increasing the signal level still further to 90 dB SPL results in a large amount of clipping in the simulated amplifier. This in turn greatly increases the measured distortion. The distortion products tend to have a flat spectrum, which is then shaped by the receiver response. The spectrum of the distortion is thus flatter than the spectrum of the linear response of instrument L20 with its 20 dB lower gain at low frequencies; this results in a greater relative amount of distortion at low frequencies than at high frequencies for this simulated hearing aid. This is illustrated in **Figures 12a** and **12b**, which show the linear hearing aid output for the 90 dB SPL input. The distortion has filled in the valley at 625 Hz for instrument L20 to a much greater degree than the valleys above 2 kHz, while the flatter response of instrument L00 results in more uniform relative distortion amounts at all frequencies. The weighted SDR index for the 90 dB SPL test signal for instrument L20 has been reduced to 0.472, and for instrument L00 the value is 0.484. One would expect some reduction in speech intelligibility under these circumstances.

The distortion measurements were then repeated for the simulated compression hearing aids, and the results are presented in **Table 10** for instrument C00, and in **Table 11** for instrument C20. The results for the 60 dB SPL input level are very similar to those for the linear hearing aids because the signal is below the compression thresholds in both channels. Increasing the signal level to 70 dB SPL results in an increase in distortion because the rapid attack time of the compression processing distorts the peaks of the test signal. The distortion can also be seen in **Figures 11c** and **11d**, where the valleys in the hearing aid output for the 70 dB SPL input contain more energy than those of **Figures 11a** and **11b** for the simulated linear instruments. Increasing the test signal level to 80 and then 90 dB SPL,

Table 8.
Ratio of distortion to signal level and the weighted SDR index as a function of the input signal intensity for the simulated linear hearing aid L00.

Frequency, Hz	Input Signal Intensity, dB SPL 60	70	80	90
625	0.08%	0.07%	5.18%	16.38%
1250	0.09	0.06	5.86	18.15
1875	0.07	0.05	5.07	15.73
2500	0.11	0.06	6.58	21.27
3125	0.12	0.06	6.35	21.85
3750	0.17	0.08	7.16	23.87
4375	0.20	0.09	7.73	26.56
5000	0.23	0.09	8.35	28.29
Index	1.000	1.000	0.821	0.484

Table 9.
Ratio of distortion to signal level and the weighted SDR index as a function of the input signal intensity for the simulated linear hearing aid L20.

Frequency, Hz	Input Signal Intensity, dB SPL 60	70	80	90
625	0.16%	0.15%	6.11%	46.32%
1250	0.08	0.07	3.01	25.54
1875	0.06	0.04	1.07	9.51
2500	0.12	0.07	1.44	12.82
3125	0.13	0.06	1.44	11.06
3750	0.17	0.08	1.54	13.18
4375	0.21	0.09	1.66	14.09
5000	0.22	0.09	1.78	15.42
Index	1.000	1.000	0.951	0.472

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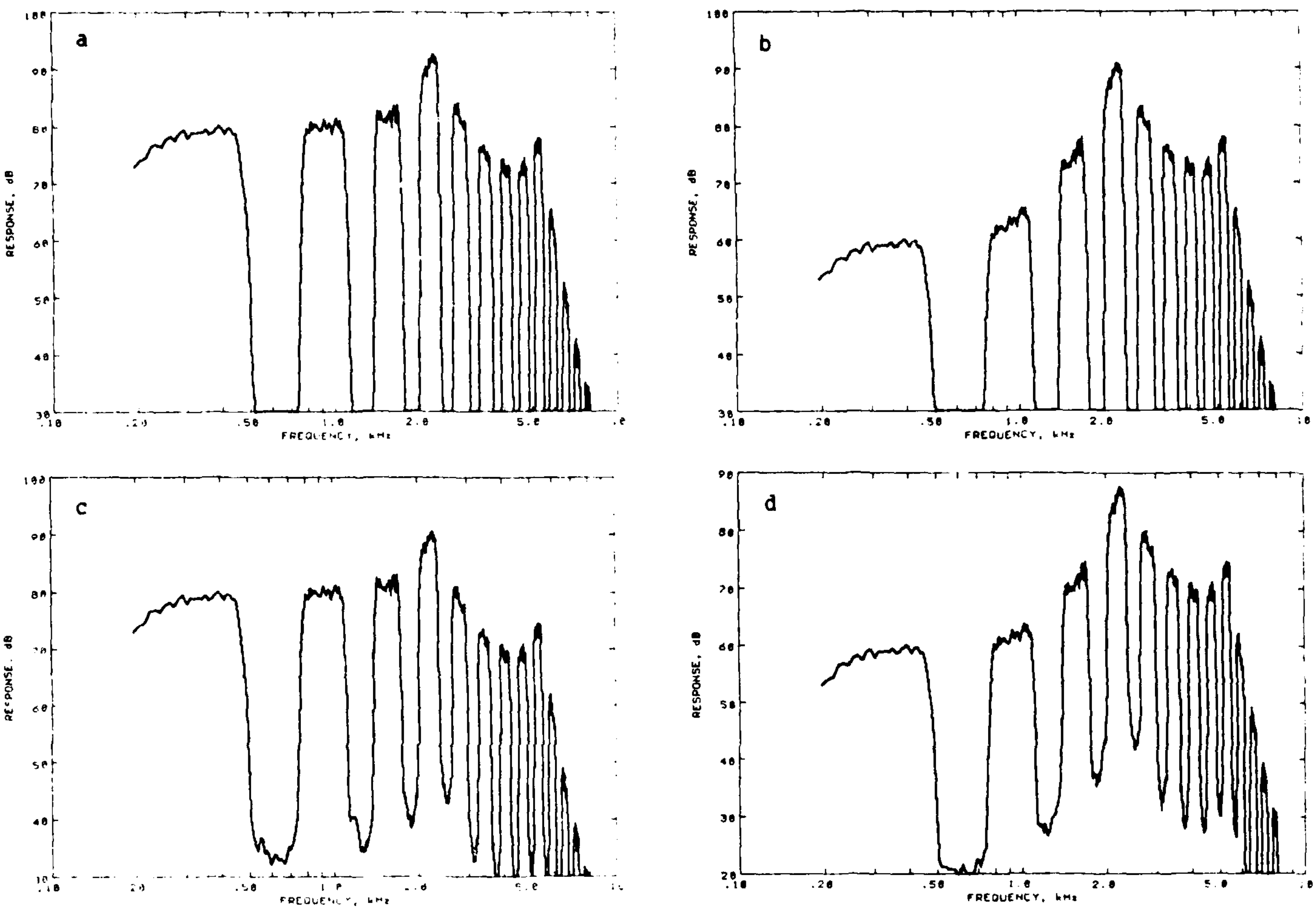

Figure 11.
Output spectrum for the comb-filtered shaped noise signal input at 70 dB SPL for: a) the simulated linear hearing aid [illegible]; b) the simulated linear hearing aid L20; c) the simulated compression aid C00; and, d) the simulated compression aid C20.

however, results in only small increases in the distortion levels because the compression prevents the peaks from exceeding the clipping threshold of the amplifier. The small amount of distortion can also be seen in the compression hearing aid frequency-response curves of **Figures 12c** and **12d** for the 90 dB SPL input signal, where the energy in the valleys is not very different from the levels for the 70 dB SPL input as shown in **Figure 11**. Also note that the weighted SDR index for the compression instrument remains at 1.000 for all the test signal levels.

Distortion measurements using the coherence function

The SDR can also be derived from the magnitude-squared coherence function using the relationship:

$$\mathrm{SDR}(f) = \frac{\gamma^2(f)}{1 - \gamma^2(f)} \qquad [1]$$

where $\gamma^2(f)$ is the magnitude-squared coherence function. Because the coherence function is a two-channel measurement procedure, processing the same amount of data as for the comb-filtered noise procedure takes twice the number of FFTs per data segment and twice the amount of processing time. Plots of the SDR were derived from the coherence function and smoothed in overlapping third-octave bands for the shaped-noise input signal ranging in level from 60 to 90 dB SPL. The data were processed in segments of 2,048 samples (102 ms) using a Hanning window and 50 percent overlap just as was done for the comb-filtered noise signal.

The SDR derived from the coherence function are presented in **Figures 13a** and **13b** for the linear instrument L20 and the compression instrument C20, respectively. The curves in **Figure 13a** for the 60 and 70 dB SPL input levels overlap for the linear hearing aid, and give a SDR of about

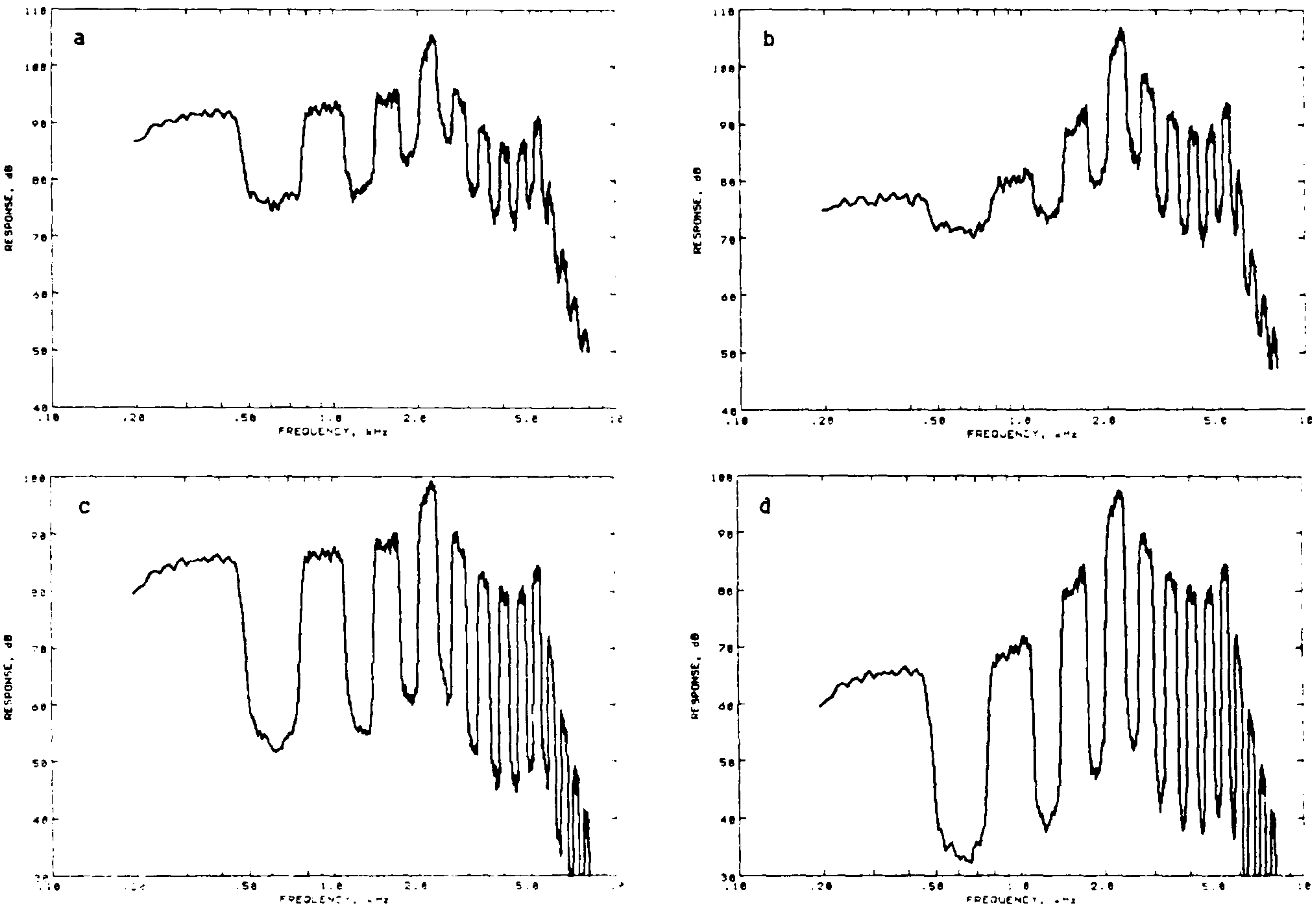

Figure 12.
Output spectrum for the comb-filtered shaped noise signal input at 90 dB SPL for: a) the simulated linear hearing aid L00; b) the simulated linear hearing aid L20; c) the simulated compression aid C00; and, d) the simulated compression aid C20.

34 dB in the vicinity of 1 kHz. This should be compared with the distortion value of 0.08 percent measured at 1250 Hz for the same linear instrument L20 using the comb-filtered noise signal, which corresponds to a SDR of about 62 dB. The SDR curves in **Figure 13a** for the 80 and 90 dB SPL input levels lie lower than the SDR curves for the lower input levels, as would be expected, and again, the SDR values are much lower than for the comb-filtered noise. The set of SDR curves is repeated in **Figure 13b** for the simulated compression instrument C20. While the trend of the distortion behavior is similar to that found with the comb-filtered noise procedure, all of the curves underestimate the SDR as compared with the comb-filtered noise measurements.

The low estimated SDR from the coherence function is the result of the bias in the estimation procedure due to truncating the data segments. Doubling the length of the data segment to 4,096 samples (205 ms) increases the estimated SDR by 6 dB, but also increases the variance since only half the number of data segments can be extracted from the same amount of test data. To increase the estimated SDR from 34 to 64 dB, while keeping the variance of the estimate constant at each frequency before smoothing, would require increasing the data segment length by a factor of 32, while keeping the number of segments the same. This would require processing 32 times the amount of data to achieve the same accuracy in the estimate as is achieved by testing with the comb-filtered noise signal. If third-octave smoothing is used, the number of data segments needed for the coherence function estimate can be reduced since the smoothing over adjacent frequency points reduces the variance, but it will still require larger FFTs and more processing time than for the comb-filtered noise procedure.

Table 10.
Ratio of distortion to signal level and the weighted SDR index as a function of the input signal intensity for the simulated compression hearing aid C00.

Frequency, Hz	Input Signal Intensity, dB SPL 60	70	80	90
625	0.08%	0.53%	2.01%	2.38%
1250	0.09	0.70	2.21	2.72
1875	0.07	0.55	2.04	2.28
2500	0.12	0.79	1.87	2.53
3125	0.13	0.89	1.54	2.30
3750	0.17	0.96	1.28	2.25
4375	0.20	1.03	1.16	2.36
5000	0.23	1.03	1.13	2.45
Index	1.000	1.000	1.000	1.000

Table 11.
Ratio of distortion to signal level and the weighted SDR index as a function of the input signal intensity for the simulated compression hearing aid C20.

Frequency, Hz	Input Signal Intensity, dB SPL 60	70	80	90
625	0.09%	1.03%	2.08%	2.15%
1250	0.07	1.17	1.47	1.40
1875	0.06	0.71	0.79	0.76
2500	0.13	0.93	0.95	0.95
3125	0.13	0.89	0.90	0.90
3750	0.17	0.92	0.92	0.92
4375	0.20	1.01	1.01	1.02
5000	0.23	1.01	1.01	1.01
Index	1.000	1.000	1.000	1.000

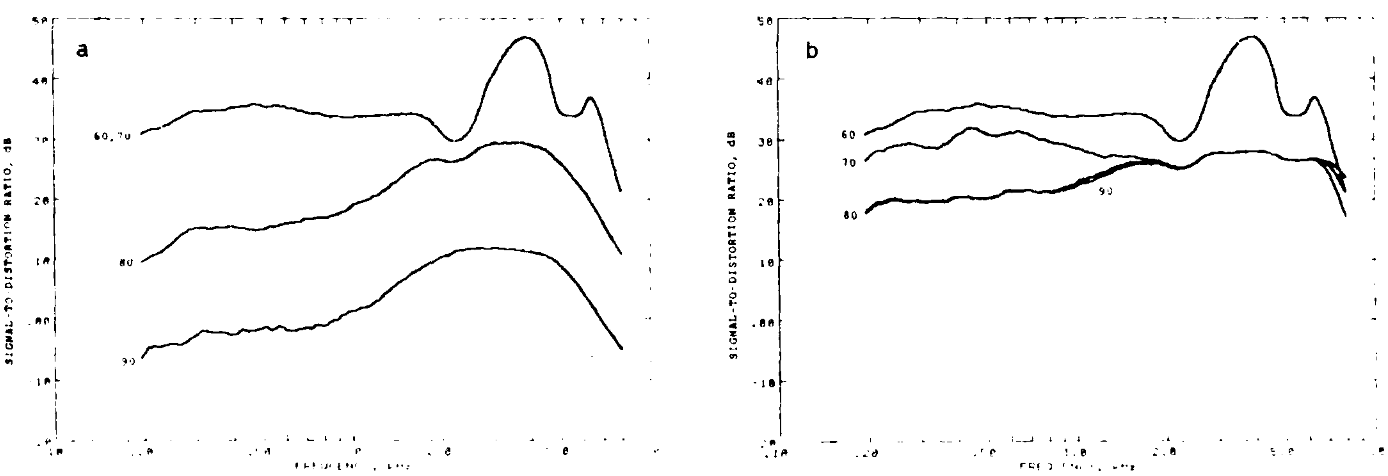

Figure 13.
Signal-to-distortion ratio derived from the magnitude-squared coherence function as the excitation goes from 60 to 90 dB SPL in steps of 10 dB for: a) the simulated linear hearing aid L20; and, b) the simulated compression hearing aid C20.

CONCLUSIONS

This paper has presented an integrated test suite for evaluating an arbitrary commercially available hearing aid. The set of tests consists of the frequency response as a function of the test signal level, the processing type and number of processing bands, input/output characteristics for each identified processing band, attack and release times for each band, and frequency-dependent broadband distortion as a function of the input signal level along with an overall weighted signal-to-distortion index. Existing test procedures have been used for the frequency-response, input/output, and attack and release time measurements. New procedures have been derived for determining the processing type and number of bands, and for measuring the broadband distortion.

The test procedure for determining the number of processing bands currently tests for only one or two bands. The procedure can be extended to a greater number of bands by increasing the length of time for the sweep to traverse the frequency range, and by increasing the number of taps in the adaptive filters. This will increase the resolution of the procedure and improve the ability to identify the band edges, but at the price of increased processing time. The set of idealized test patterns can then be extended to include three or more channels of independent compression or any other processing or combinations of processing to be checked.

The test suite is suitable for the automated testing of a hearing aid. The number of processing bands and the band edges determine the frequencies to be used for testing the input/output characteristics and the attack and release times, so a simple algorithm can step through the five sets of tests presented in this paper in the order presented and produce a set of test results at the end.

The test suite has been designed for implementation on a personal computer containing a A/D and D/A converter board for the signal generation and data acquisition. Such a computer system is being designed at the Lexington Center for measuring actual hearing aids. Test results from this implementation will be available at a later date, and will allow the comparison of the test procedures presented in this paper with other measurement protocols.

ACKNOWLEDGMENT

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APPENDIX A: SHAPED NOISE SIGNAL

The broadband noise stimulus consists of white Gaussian noise passed through a high-pass filter at 200 Hz and a low-pass filter at 5000 Hz to restrict the bandwidth, and a one-pole low-pass filter at 900 Hz to give a speech-shaped spectrum. The high-pass and low-pass bandlimiting digital filters are derived from 3-pole analog Butterworth filters, the high-pass via the bilinear transformation and the low-pass via impulse invariance, and the one-pole analog filter at 900 Hz is also mapped into a digital filter via impulse invariance.

Define the analog filter frequencies in Hz:

f_s = sampling rate (20000 Hz)
f_h = high-pass limiting frequency (200 Hz)
f_l = low-pass limiting frequency (5000 Hz)
f_p = low-pass speech-shaping frequency (900 Hz).

The prewarped digital cutoff frequency for the high-pass filter is then

$$\alpha = \tan(\pi f_h / f_s). \qquad [A1]$$

The digital high-pass limiting filter (at 200 Hz) is then, from the bilinear transformation:

$$H(z) = \frac{1 - 3z^{-1} + 3z^{-2} - z^{-3}}{a_0 + a_1 z^{-1} + a_2 z^{-2} + a_3 z^{-3}} \qquad [A2]$$

where

$$\begin{aligned} a_0 &= (1 + \alpha)(1 + \alpha + \alpha^2) \\ a_1 &= (1 + \alpha)(2\alpha^2 - 2) + (\alpha - 1)(1 + \alpha + \alpha^2) \\ a_2 &= (\alpha - 1)(2\alpha^2 - 2) + (\alpha + 1)(1 - \alpha + \alpha^2) \\ a_3 &= (\alpha - 1)(1 - \alpha + \alpha^2). \end{aligned}$$

In the time-domain, the filter becomes

$$\begin{aligned} y(n) = [&x(n) - 3x(n-1) + 3x(n-2) - x(n-3) \\ &- a_1 y(n-1) - a_2 y(n-2) - a_3 y(n-3)] / a_0 \end{aligned} \qquad [A3]$$

where x(n) is the filter input and y(n) is the filter output.

The digital low-pass limiting filter (at 5000 Hz) is then, via impulse invariance

$$L(z) = g \left[\frac{1}{1 - b_0 z^{-1}} + \frac{-1 + b_1 z^{-1}}{1 - b_2 z^{-1} + b_0 z^{-2}} \right] \qquad [A4]$$

where

$$\begin{aligned} b_0 &= \exp(-2\pi f_l / f_s) \\ b_1 &= \exp(-\pi f_l / f_s) \times \\ &\quad [\cos(\sqrt{3}\ \pi f_l / f_s) + \sin(\sqrt{3}\ \pi f_l / f_s) / \sqrt{3}\ ] \\ b_2 &= 2 \exp(-\pi f_l / f_s) \cos(\sqrt{3}\ \pi f_l / f_s). \end{aligned}$$

The transfer function is normalized by g to get unity gain at low frequencies. In the time-domain, this filter is represented by

$$\begin{aligned} y_1(n) &= g\ x(n) + b_0 y_1(n-1) \\ y_2(n) &= -g\ x(n) + g\ b_1 x(n-1) + b_2 y_2(n-1) - b_0 y_2(n-2) \\ y(n) &= y_1(n) + y_2(n) \end{aligned} \qquad [A5]$$

where x(n) is the filter input and y(n) is the filter output.

The low-pass filter for shaping the speech spectrum is given by

$$G(z) = \frac{\gamma_0}{1 - \gamma_1 z^{-1}}$$

where

$$\begin{aligned} \gamma_0 &= 1 - \exp(-2\pi f_p / f_s) \\ \gamma_1 &= 1 - \gamma_0 \end{aligned} \qquad [A6]$$

In the time-domain, this filter becomes

$$y(n) = \gamma_0 x(n) + \gamma_1 y(n-1) \qquad [A7]$$

where x(n) is the filter input and y(n) is the filter output.

APPENDIX B: SWEPT SINUSOID

An assumption in testing a multiband hearing aid is that the frequency bands will tend to be uniformly spaced on a logarithmic frequency scale (typical of analog filter designs or approximations to auditory critical bands) as opposed to a linear frequency scale (typical of digital filter designs). The swept sinusoid should therefore spend equal amounts of time within equal logarithmic frequency bands.

Let

N = duration of the sweep (53248 samples)
f_0 = starting frequency for the sweep (222 Hz)
f_1 = stopping frequency for the sweep (4470 Hz)
f_s = sampling frequency (20000 Hz).

The desired instantaneous frequency is given by

$$\omega(n) = 2\pi(f_0 / f_s) \exp(\zeta n) \qquad [B1]$$

where

$$\zeta = \frac{1}{N-1} \ln\left(\frac{f_1}{f_0}\right).$$

The digital chirp signal for the frequency sweep is then

$$s(n) = \sin[\omega(n) / \zeta],\ 0 \le n \le N-1. \qquad [B2]$$

APPENDIX C: ADAPTIVE CANCELLATION

The adaptive cancellation system uses a pair of filters, each having N taps, as shown in **Figure 2**. Let s(n) be the hearing aid response to the sweep alone, which is the reference input to adaptive filter w(n). Let r(n) be the hearing aid response to the shaped noise alone, which is the reference signal input to

adaptive filter v(n). The hearing aid response to the sweep plus noise is denoted by x(n), and the response delayed by N/2 samples is $\hat{x}(n)$. The error signal is given by e(n), and the system adapts to drive e(n) as close to zero as possible.

The power of the sweep-plus-noise signal changes with time because of the frequency dependent gain of the hearing aid. In order to maintain stability in the adaptive system, the adaptation time constant has been made dependent on the estimated signal power (11). The estimated signal power is given by

$$\sigma^2(n) = \beta\, \sigma^2(n-1) + (1-\beta)\, \hat{x}^2(n) \qquad [C1]$$

where

$$\beta = \exp(-1 / 200)$$

for the sampling rate of 20 kHz. The adaptation time constant is then given by

$$\mu(n) = \frac{1}{4\, N\, \sigma^2(n)} \qquad [C2]$$

where N is the adaptive filter length in samples. A value of N = 15 was used for the processing examples presented in this paper.

The error signal is given by

$$e(n) = \hat{x}(n) - \sum_{k=0}^{N-1} w_k(n)\, s(n-k) - \sum_{k=0}^{N-1} v_k(n)\, r(n-k) \qquad [C3]$$

and the updating of the adaptive filter coefficients (9) is then given by

$$w_k(n) = w_k(n-1) + 2\,\mu(n)\, e(n)\, s(n-k) \qquad [C4a]$$

and

$$v_k(n) = v_k(n-1) + 2\,\mu(n)\, e(n)\, r(n-k). \qquad [C4b]$$

APPENDIX D: COMB FILTER

The shaped noise is convolved with a comb filter to produce the notched distortion test signal. Let M be the number of valleys (or peaks) in the comb filter between 0 Hz and $f_s/2$, the highest frequency that can be represented in the digital system. The filter was designed with M = 16. The comb is uniformly spaced on a linear frequency scale, and has unity gain for the peaks of the comb and −62 dB gain for the valleys. The comb filter, designed using the Parks-McClellan algorithm (16), is given by

$$H(z) = \sum_{k=0}^{30} h_k\, z^{-2Mk} \qquad [D1]$$

where

$$\begin{aligned}
h_0 &= 0.1476714 \times 10^{-2} = h_{30}\\
h_1 &= 0 = h_{29}\\
h_2 &= -0.3777827 \times 10^{-2} = h_{28}\\
h_3 &= 0 = h_{27}\\
h_4 &= 0.8223298 \times 10^{-2} = h_{26}\\
h_5 &= 0 = h_{25}\\
h_6 &= -0.1578323 \times 10^{-1} = h_{24}\\
h_7 &= 0 = h_{23}\\
h_8 &= 0.2832202 \times 10^{-1} = h_{22}\\
h_9 &= 0 = h_{21}\\
h_{10} &= -0.5025136 \times 10^{-1} = h_{20}\\
h_{11} &= 0 = h_{19}\\
h_{12} &= 0.9755109 \times 10^{-1} = h_{18}\\
h_{13} &= 0 = h_{17}\\
h_{14} &= -0.3153704 = h_{16}\\
h_{15} &= 0.5000000.
\end{aligned}$$

The comb filter in the time-domain is given by

$$y(n) = \sum_{k=0}^{30} h_k\, x(n-2Mk) \qquad [D2]$$

where y(n) is the output of the filter and x(n) the input. For M = 16, the total filter length is 961 samples, of which only 17 are non-zero.

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A time-domain digital simulation of hearing aid response

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Abstract—A time-domain digital simulation of an in-the-ear (ITE) hearing aid has been developed. The simulation allows modeling of nonlinear effects such as compression and amplifier distortion in addition to linear processing and acoustics. The simulation includes a microphone, two-channel compression processing, an amplifier with clipping distortion, a receiver, an ear canal and ear drum, and feedback and direct sound transmission through the vent. Simulation results for a linear hearing aid are similar to those obtained for frequency-domain representations of the analog system. Examples of responses for nonlinear systems are also provided.

Key words: *amplifier, analog system, in-the-ear (ITE) hearing aid, microphone, time-domain digital simulation, tone-burst responses, two-channel compression processing.*

INTRODUCTION

This paper presents a time-domain digital simulation of an in-the-ear (ITE) hearing aid. Previous computer simulations, such as those of Egolf, *et al.* (4), Bade, *et al.* (1), and Kates (5) have all been in the frequency domain and limited to representing the linear behavior of a hearing aid. The time-domain simulation, however, extends the modeling to include nonlinear effects such as compression and amplifier distortion. This results in a more complete computer simulation of a hearing aid than previously available, and one which can be used for many applications such as the development of hearing aid signal processing and hearing aid test systems.

A time-domain model is essential for simulating the effects of nonlinear or linear time-varying systems in which the processing changes in response to the input signal behavior. Examples in hearing aids include compression in one or more frequency channels, compression with adaptive time constants, adaptive filters, noise suppression, and feedback cancellation. The design and evaluation of such algorithms can be greatly facilitated by using a computer simulation during the processing development. Distortion is also of interest; a common example is amplifier saturation. Modeling the distortion characteristics of typical circuits makes it possible to isolate the effects of distortion on sound quality and speech intelligibility in hearing aids.

The time-domain simulation is based on the frequency-domain simulation of Kates (5). That simulation includes the linear effects of a microphone, amplifier, receiver, vent, ear canal and ear drum, and pinna. The results of the frequency-domain model were shown by Kates (5) to be accurate in reproducing the measured behavior of an ITE hearing aid mounted on the KEMAR anthropometric manikin (2) and to be useful in predicting the effects of varying the acoustics of the hearing aid and the ear (5,6). The purpose of the time-domain simulation is not, however, to reproduce the behavior of a specific hearing aid, but rather to simulate a generic hearing aid incorporating signal processing of interest.

In developing the time-domain simulation, most of the elements of the frequency-domain model of Kates (5) were kept by converting the frequency response of each element in the model into a corresponding time-domain digital filter

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response. The pinna effects, which convert the free-field excitation into the pressure at the microphone, have not been included because they are equivalent to a modification of the excitation signal and can be added at a later date if needed. The amplifier, instead of modeling a specific manufacturer's hearing aid, has been replaced by an ideal voltage or current source with an adjustable hard-clipping threshold to simulate an amplifier with distortion.

Because the simulated amplifier does not duplicate the behavior of any specific hearing aid, validation of the model is based on general behavior and the correspondence with earlier models rather than with matching a set of individual hearing aid temporal responses. Hearing aid processing has been added to the model to give a two-channel compression hearing aid with independent input-referred automatic gain control (AGC-I) in each channel and an adjustable crossover frequency between the two channels.

The purpose of this paper is to describe the time-domain simulation and to give some examples of its use. The underlying physics of the model has been treated by Kates (5) and is therefore not repeated here. The examples show the frequency response of a simulated linear system with acoustic feedback computed from the system impulse response; the temporal response of a simulated hearing aid to a 1-kHz tone burst for three processing options (ideal linear, the same instrument with clipping distortion, and two-channel compression); and the response of the three processing systems to a two-tone test signal used for the measurement of intermodulation distortion.

ELEMENTS OF THE SIMULATION

The computer program simulates a standard (full-concha) ITE hearing aid. The major external features of the hearing aid are the microphone opening, battery compartment, volume control, vent openings, and receiver opening. A cross-section of the typical ITE hearing aid is presented in **Figure 1**. The microphone is positioned near the top of the hearing aid faceplate above the battery compartment and the volume control; the vent is at the bottom. Not shown is circuitry internal to the instrument; it is positioned where there is available space. The receiver is located in the canal portion of the hearing aid, and the receiver output is conducted to the ear canal via a short tube. The vent runs from the faceplate to the ear canal and can take several shapes, although a tube having circular cross-section is the most common and is a reasonable acoustic model for other geometries as well. The hearing aid fits entirely in the ear: the portion containing the

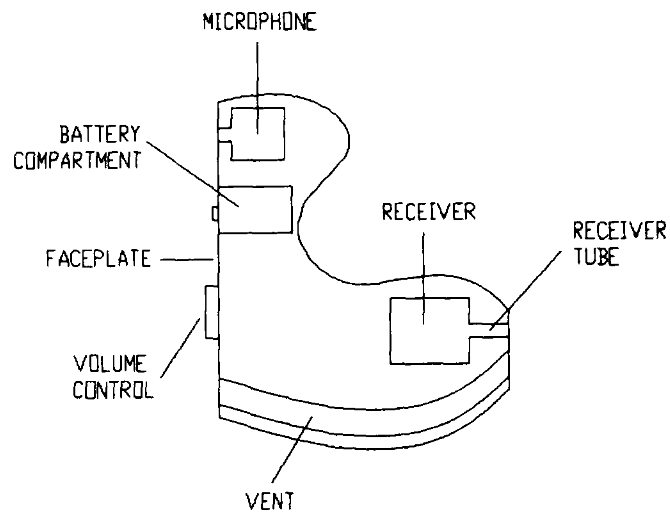

Figure 1.
Cross-section of an in-the-ear (ITE) hearing aid.

receiver is inserted in the ear canal and the portion containing the microphone and battery compartment rests in the concha area of the outer ear.

A block diagram of the simulated instrument for 2 cm^3 coupler measurements is presented in **Figure 2**. The input to the microphone is the free-field sound pressure generated by an ideal loudspeaker because no pinna or head effects are present. The microphone output is split into a low-pass and a high-pass channel, with independent input-referred compression (AGC-I) in each channel. The gain in each channel is then adjusted; following this, signals are summed. The power-amplifier behavior is represented as a symmetric clipping stage followed by a fixed gain. The receiver is assumed to be loaded acoustically by a tube leading to a 2 cm^3 coupler; any vent present is ignored because it is not connected to the coupler for this type of measurement.

A block diagram of the simulated instrument inserted into an ear is presented in **Figure 3**. The vent now becomes part of the system. In addition to the amplified signal path shown in **Figure 2**, there is also an unamplified signal path directly through the vent, so the sound pressure in the ear canal is the sum of the amplified and direct signals. Acoustic feedback is included in the model as sound from the ear canal being transmitted through the vent to the faceplate where it is reradiated and picked up by the microphone. The receiver, connected to the ear canal by a short tube, is loaded by tube, vent, and ear canal. The ear canal is terminated by the modified Zwislocki coupler used in the KEMAR manikin. The digital time-domain simulation has been implemented at a sampling rate of

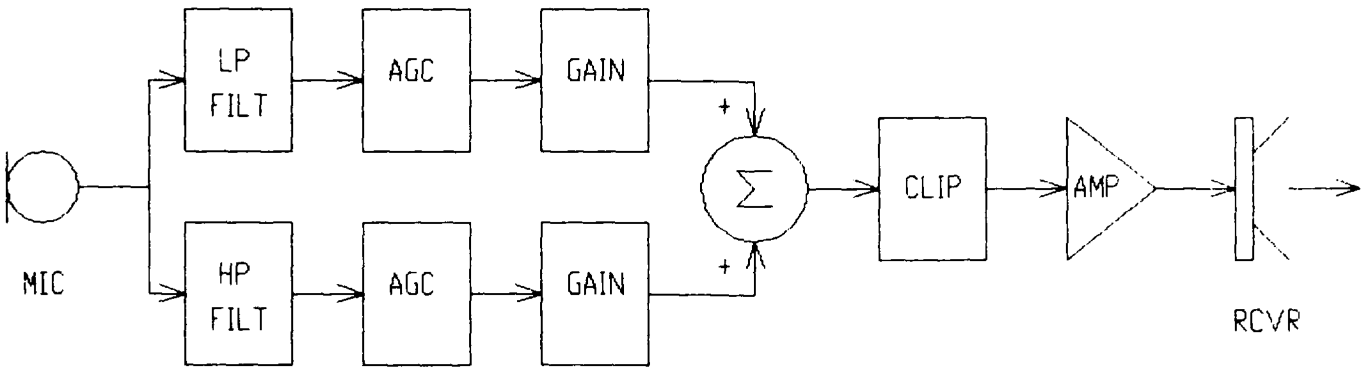

Figure 2.
Block diagram of the simulated hearing aid for 2 cm^3 coupler measurements.

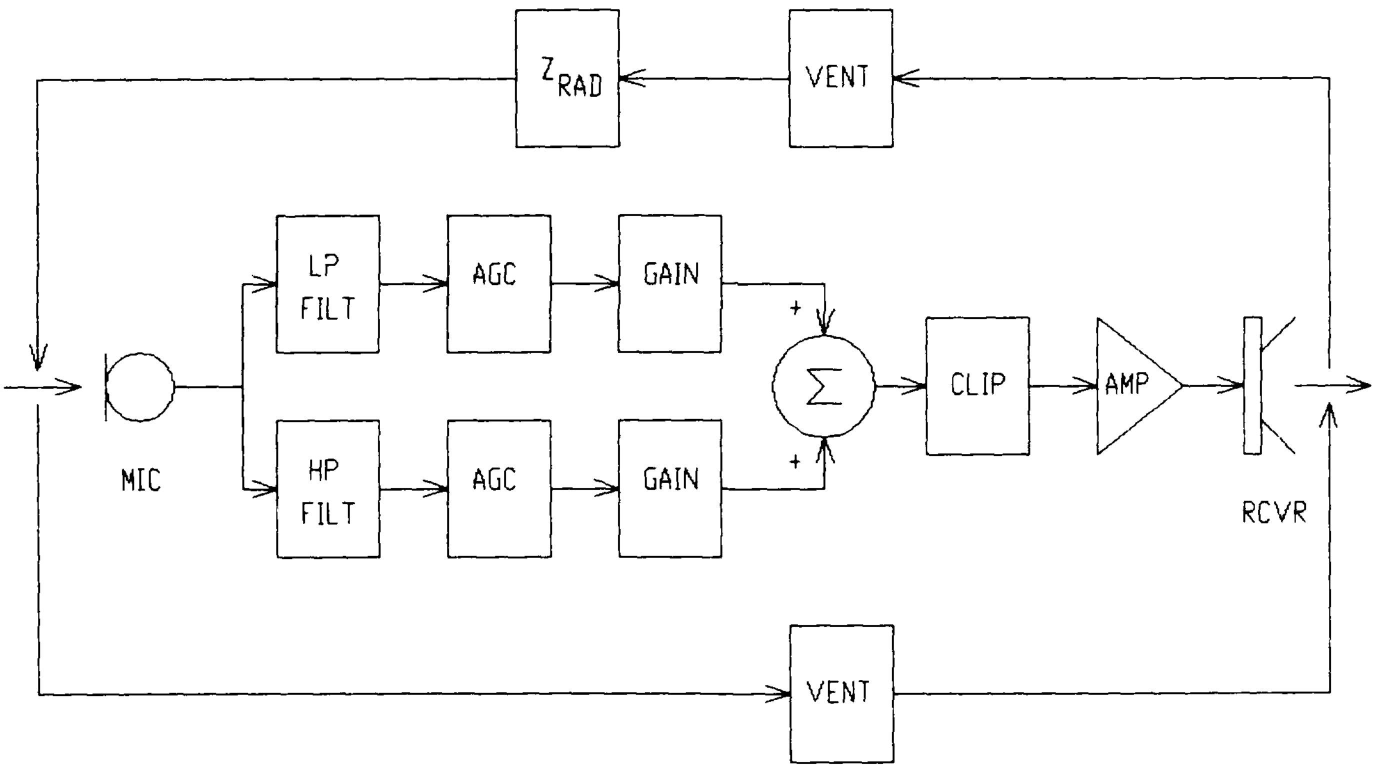

Figure 3.
Block diagram of the simulated hearing aid with vent.

20 kHz using the elements described in the following paragraphs.

Microphone

The microphone is a digital transformation of the transfer function of a Knowles EA-1842 microphone as modeled by Kates (5). The microphone is represented as a one-pole high-pass filter at 300 Hz in cascade with a two-pole low-pass filter at 5100 Hz having a Q of 1.55. The digital high-pass filter was implemented using the bilinear transformation (7) and the digital low-pass filter was implemented using impulse invariance (7). The frequency response of the digital microphone simulation is shown in **Figure 4**. Note that the microphone response has been adjusted to give 0 dB gain at 1 kHz.

Filters

The low-pass and high-pass filters are three-pole Butterworth filters. The digital filters are bilinear transformations of the analog transfer functions, with the crossover

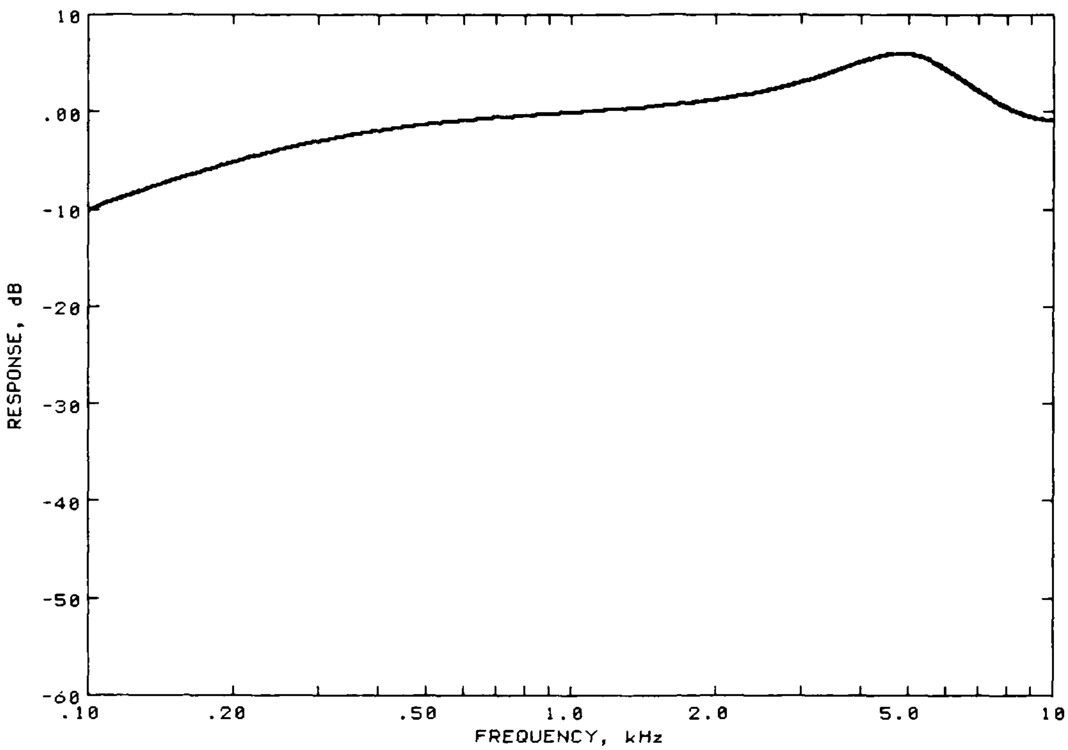

Figure 4.
Transfer function for the simulated Knowles EA-1842 microphone.

frequency a parameter of the hearing aid simulation. The digital filter responses are shown in **Figure 5** for a crossover frequency of 1 kHz. One property of these filters is that the sum of the low-pass and high-pass filters gives a smooth frequency response for any setting of the relative gains of the two filters. Other filter designs may not share this feature. The overall response is illustrated in **Figure 6**, where the gain of the high-pass filter is increased from 0 dB to 30 dB in steps of 10 dB, while the gain of the low-pass filter is kept at 0 dB. The combination of the two filters yields an overall frequency response that monotonically increases with increasing frequency.

AGC-I

The two frequency channels have identical independent AGC-I circuits. The analog prototype peak-detection circuit used in the AGC is shown in **Figure 7**. The circuit input is the absolute value of the low-pass or high-pass filter output. The attack time constant is given by $\tau_1 = R_1C$, under the assumption that the attack time is much shorter than the release time. The release time constant is given by $\tau_2 = R_2C$. The peak voltage v(t) is used to control the AGC gain for the channel. Circuit equations were transformed using backwards differences (7) to give the digital simulation, with the attack and release times in each channel parameters of the model.

The input/output behavior of each AGC circuit is shown in **Figure 8**. The system is linear as long as the control voltage v(t) is below the AGC threshold. Compression is engaged when the control voltage exceeds the threshold, with the threshold and the compression ratio a parameter of the simulation for each channel. The compression behavior is therefore governed by the estimated peak input signal levels. The compression threshold is specified in terms of the equivalent input sound-pressure level in dB sound-pressure level (SPL) at 1 kHz, so the actual compression threshold is modified by the microphone response of **Figure 4** and the low-pass or high-pass filter response from **Figure 5**.

Amplifier

The gain of the simulated system is specified for each frequency channel separately; following this, the signals are combined. The simulated amplifier then converts this signal into a form suitable for driving the receiver. The amplifier is modeled as a voltage or current source with adjustable clipping distortion; a voltage source is commonly used in high-power hearing aids and a current source is commonly used in low- to moderate-power hearing aids. The amplifier output impedance is either zero ohms for

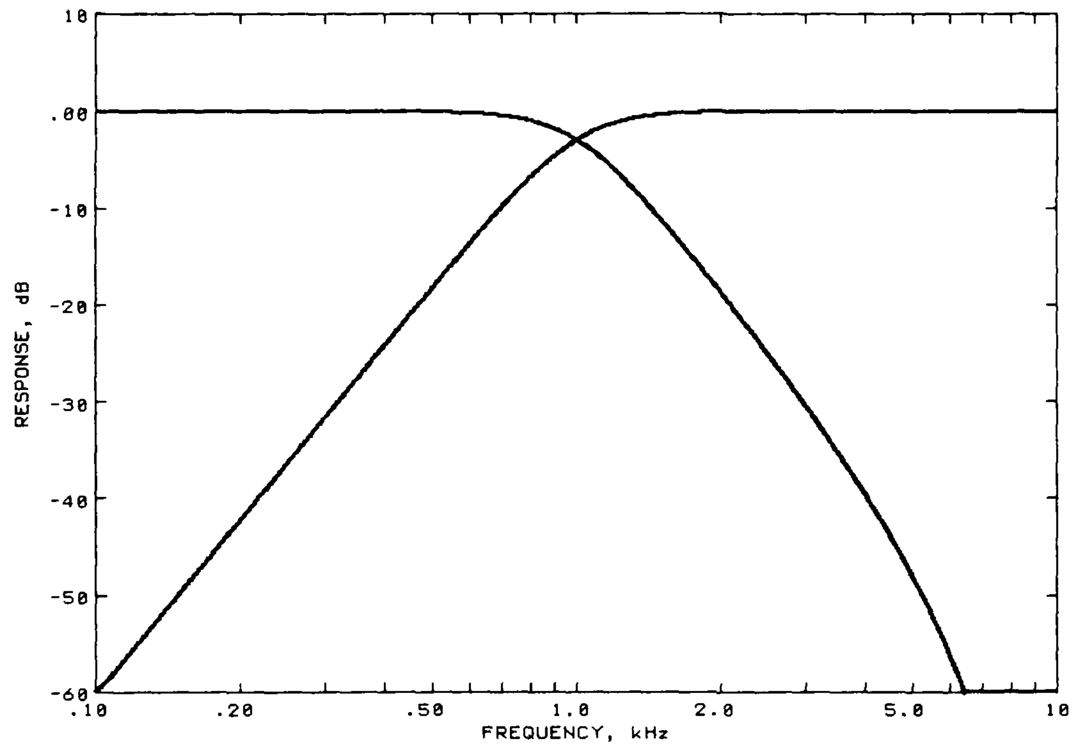

Figure 5.
Low-pass and high-pass filter frequency responses for the simulated hearing aid.

the voltage source or 100,000 ohms for the current source. The amplifier distortion is simulated as symmetric clipping of either the voltage or the current, depending on the type of source selected.* The clipping level is specified in terms of the equivalent input peak SPL at 1 kHz with 0 dB gain in both frequency channels, so the actual clipping level is modified by the microphone response and the gain in each frequency channel.

Receiver

The digital receiver simulation, based on the electrical analog of Carlson (3), was implemented by Kates (5) for the frequency-domain simulation. The output impedance of the amplifier is included in the receiver model, as is the specified acoustic load of ear canal, ear drum, and vent. The resultant receiver equivalent circuit model is very complicated, so a frequency-sampling approach (7) was used to generate the time-domain response instead of an analytical technique such as the bilinear transformation.

To compute the time-domain simulation for the sampling rate of 20 kHz, the complex analog frequency-domain transfer function was evaluated at 512 points uniformly spaced from 0 to 10 kHz. This frequency response was transformed into the time-domain via an inverse fast Fourier transform (FFT) to give a 1,024-point impulse response. The number of samples was chosen to minimize aliasing in the impulse response while keeping a reasonable computational burden. The impulse response was then windowed to 256 samples using the function given below:

$$w(n) = \begin{cases} 1 & 0 \leq n \leq 127 \\ 0.5\left[1 + \cos(\pi \frac{n-128}{128})\right] & 128 \leq n \leq 255 \\ 0 & 256 \leq n \leq 1023 \end{cases} \quad [1]$$

The windowed sequence yields a finite impulse response (FIR) digital approximation to the analog system. The amplifier output signal is then convolved with the FIR receiver response to give the receiver output.

The resultant frequency response for a Knowles ED-1913 receiver connected to a 2 cm^3 coupler is shown in **Figure 9**. The solid line is the frequency response computed from the frequency-domain model of the analog system. The dashed line is the Fourier transform of the windowed digital impulse response. The pressure measurement point is at the junction of the receiver tube with the ear canal at the end of the hearing aid. The agreement between the two curves is quite good since they overlap over almost the entire frequency range, thus showing the

*Personal communication, S. Armstrong, Gennum Corp., 1989.

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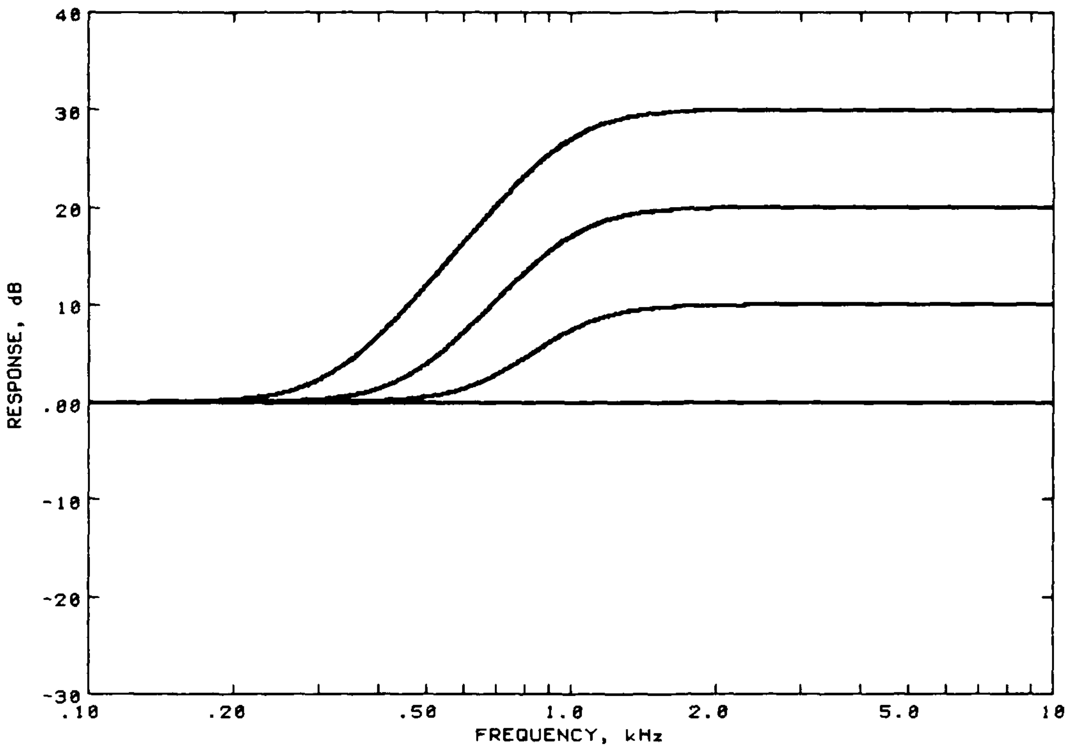

Figure 6.
Summed filter outputs as the gain of the high-pass filter channel is increased from 0 dB to 30 dB in steps of 10 dB. The gain of the low-pass filter channel is 0 dB.

accuracy of this simulation approach.

Figure 10 shows the frequency response for the same receiver loaded with a simulated ear and vent, but with the feedback and feedforward signal paths through the vent left unconnected. The parameters for the ear and vent are given in **Table 1**; these values are similar to those used by Kates (5,6) in simulating an actual vented hearing aid. The solid curve in the figure is again the frequency response from the frequency-domain simulation; the dashed line is the Fourier transform of the windowed impulse response. There is some minor discrepancy between the two curves in the low-level response regions, approximately 40 dB below the peaks, due to the effects of the window and the sampling rate. Such deviations will occur for any transformation of an analog system into the digital domain. The agreement at high response levels, however, is still very good. In particular, the effect of the vent at low frequencies, where the mass of the air in the vent and the compliance of the air in the ear canal interact to produce a high-pass filter, are quite apparent in the peak at 600 Hz and the reduction in response below that frequency. Because there is no feedback for this response calculation, the effect at high frequencies is primarily a notch at 7 kHz caused by the vent half-wave resonance modifying the acoustic load on the receiver.

Table 1.
Acoustic parameters of the simulated ear and hearing aid.

Receiver termination model	
Type:	tube
Length:	0.6 cm
Radius:	0.06 cm
Ear canal model (unoccluded portion)	
Type:	tube
Length:	1.2 cm
Radius:	0.33 cm
Ear-drum impedance:	modified Zwislocki coupler
Vent model	
Type:	tube
Length:	2.2 cm
Radius:	0.12 cm
Vent-Mic distance:	1.8 cm

Vent

The vent carries sound in both directions, as shown in the block diagram of **Figure 3**. It requires two different digital simulation responses because the acoustic source

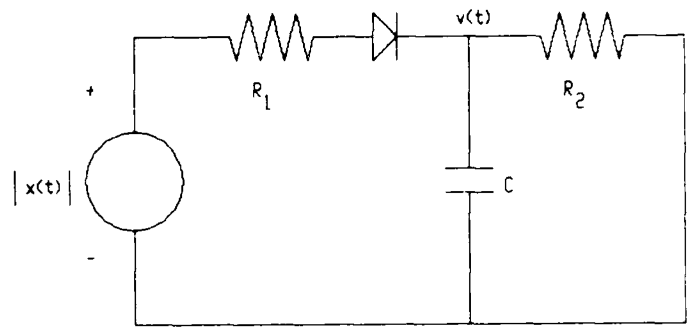

Figure 7.
Peak-detection circuit simulated in each channel of the hearing aid.

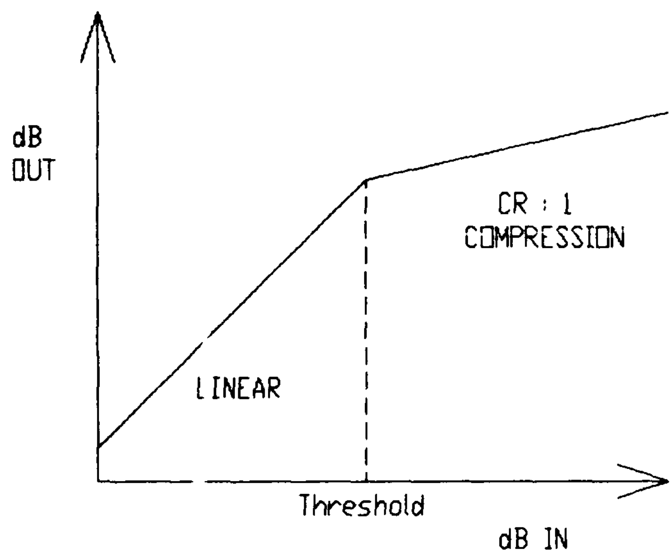

Figure 8.
Input/output characteristics of each channel in the simulated hearing aid.

and load impedances are different for the two transmission paths through the vent (5). The vent responses were computed using the same procedure as used for the receiver response.

The feedback-path frequency response, including the radiation impedance at the vent opening in the faceplate and the propagation from the vent to the microphone, is shown in **Figure 11** for the vent and ear parameters given in **Table 1**. The solid line is the frequency response computed from the frequency-domain simulation; the dashed line is the Fourier transform of the windowed impulse response. The agreement between the two curves is quite good, with both showing an overall gain of about −56 dB at low frequencies. The feedback path has a sharp resonance peak near 7 kHz. At this frequency, the effective length of the vent loaded by the radiation impedance is approximately a half wavelength.

The peak in the vent feedback response at 7 kHz does not mean that the hearing aid will tend to go into oscillation at this frequency. Oscillation is symptomatic of an unstable system, in which the amplitude and phase of the feedback path combine to give nearly unity gain. The vent response near 7 kHz is compensated by the low receiver output in the same frequency region, and the overall phase shift results in a notch rather than a peak for the system represented in **Figure 11**. In most cases, oscillation occurs at frequencies in the vicinity of the receiver response peaks since this is where the instrument has the highest gain. The exact frequency will depend on the phase shifts as well as the amplitudes of the amplified and feedback signal paths.

The direct-path response for the same simulated vent and ear is shown in **Figure 12**. The basic behavior of the direct path through the vent is a low-pass filter. The resonance frequency is determined by the interaction of the mass of air in the vent and the compliance of the air in the ear canal, and the sharpness of the resonance affected in addition by the losses in the vent and ear. The vent response also shows the high-frequency resonance, but the resonance frequency has moved to about 7.5 kHz due to the difference in the vent load impedance. Again, the frequency response computed from the windowed impulse response (dashed line) is in good agreement with the frequency-domain response (solid line).

EXAMPLES

Frequency response

The frequency response of the simulated linear system was obtained from the Fourier transform of the system impulse response for a hearing aid with and without a vent. The parameters of the simulated linear instrument are given in **Table 2**. The parameters for the ear are from **Table 1**, and the vent (when included) is also specified in **Table 1**. The time-domain simulation was run at a sampling rate of 20 kHz.

The frequency response of the hearing aid without the vent is given by the dashed line in **Figure 13**. Because both the low-pass and the high-pass processing channel have the same gain, the overall magnitude frequency response of the simulated hearing aid is that of the microphone plus receiver. This curve is quite similar to the unvented response curves computed by Kates (5) using the frequency-domain simulation for the same microphone, receiver, and model parameters, but with a different simulated amplifier.

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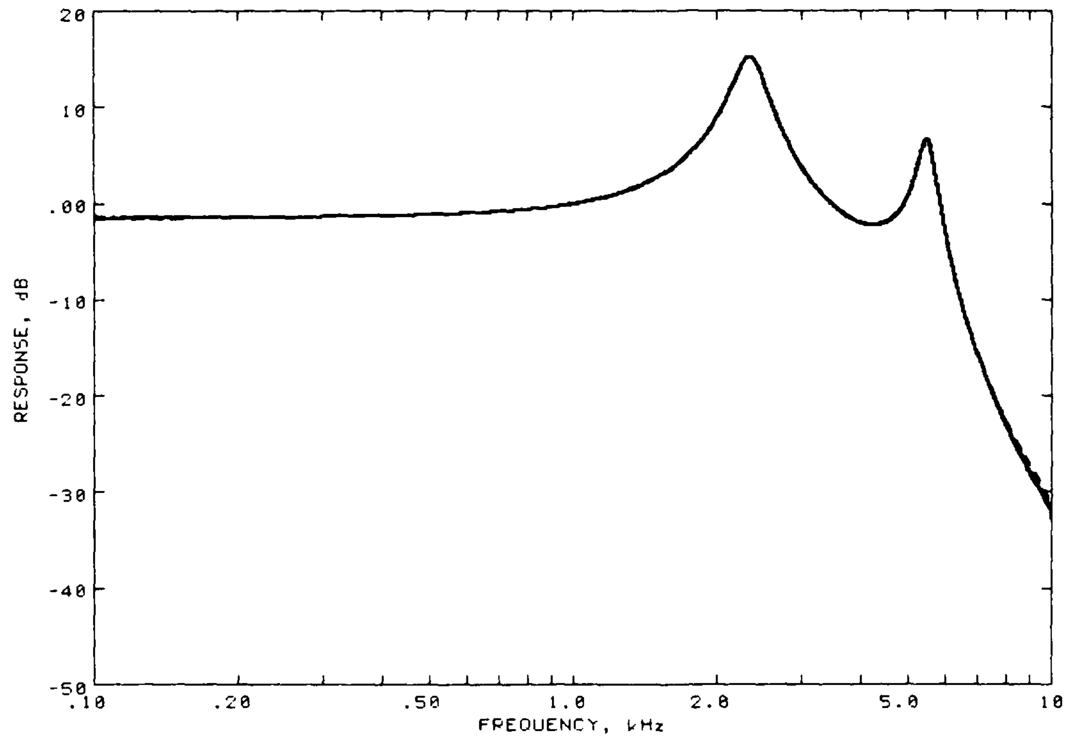

Figure 9.
Coupler pressure transfer function for the simulated Knowles ED-1913 receiver terminated with a tube 0.6 cm long and having a radius of 0.06 cm leading to the 2 cm^3 coupler. The solid line is the target frequency response, the dashed line the simulation.

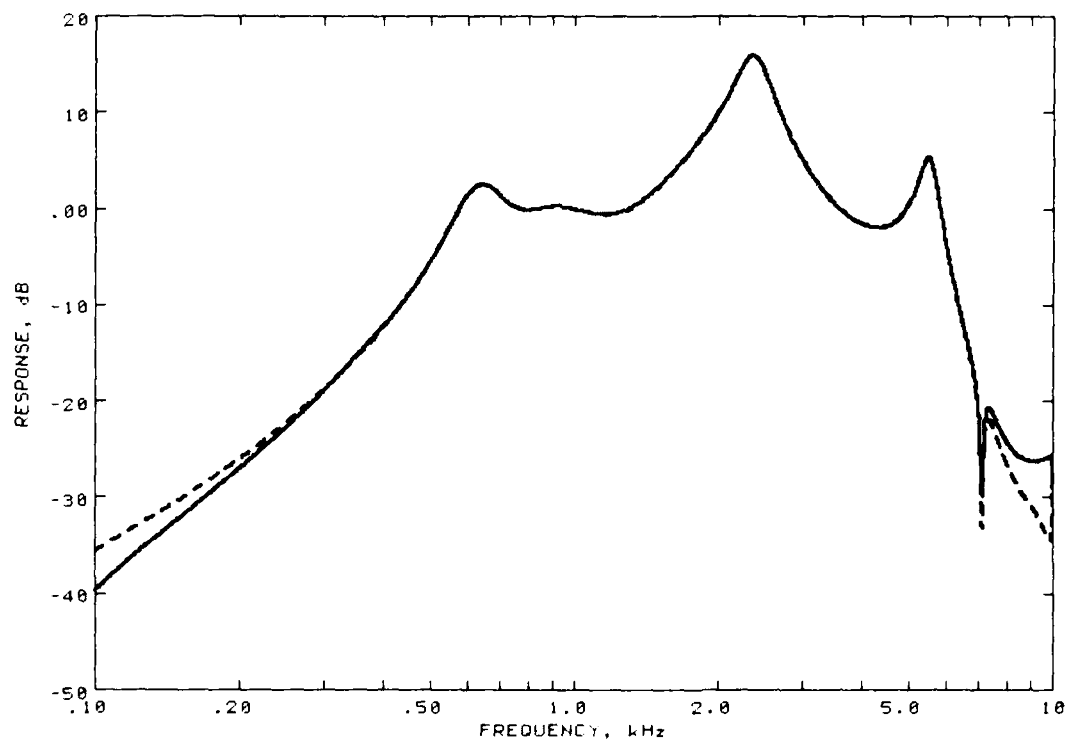

Figure 10.
Pressure transfer function for the simulated Knowles ED-1913 receiver terminated with the tube, vent, and ear canal specified in **Table 1.** Pressure is measured at the tube opening into the ear canal. The solid line is target frequency response, dashed line the simulation.

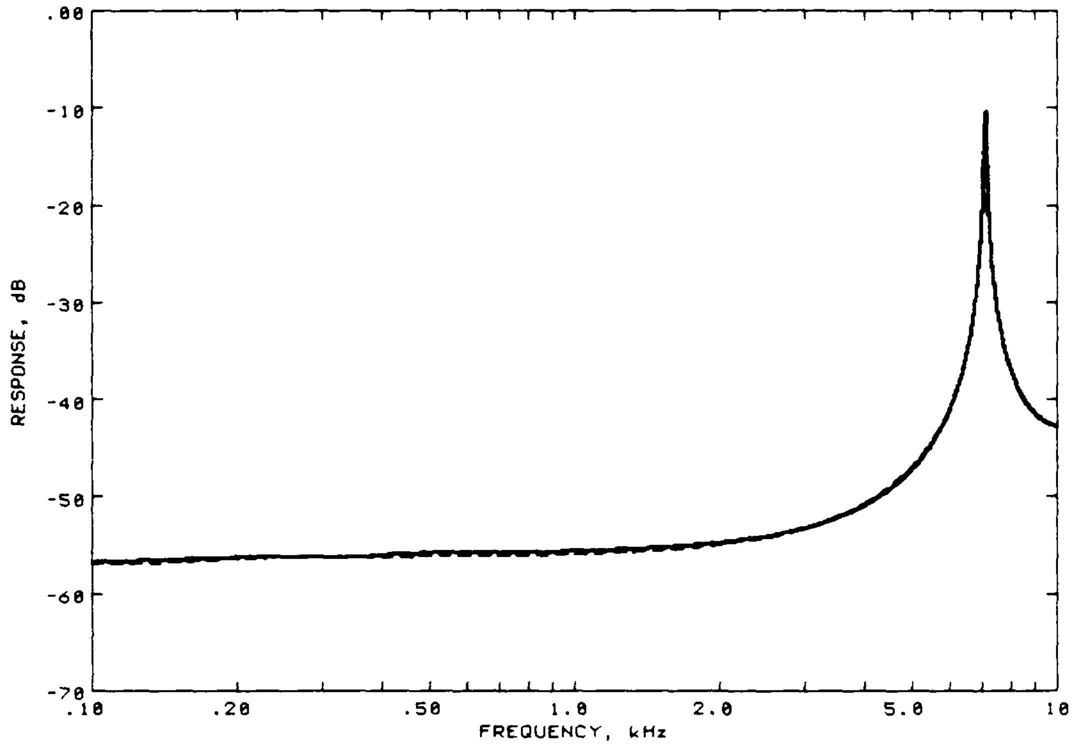

Figure 11.
Pressure transfer function for the feedback path through the vent and back to the microphone for the vent specified in **Table 1**. The solid line is the target response, the dashed line the simulation.

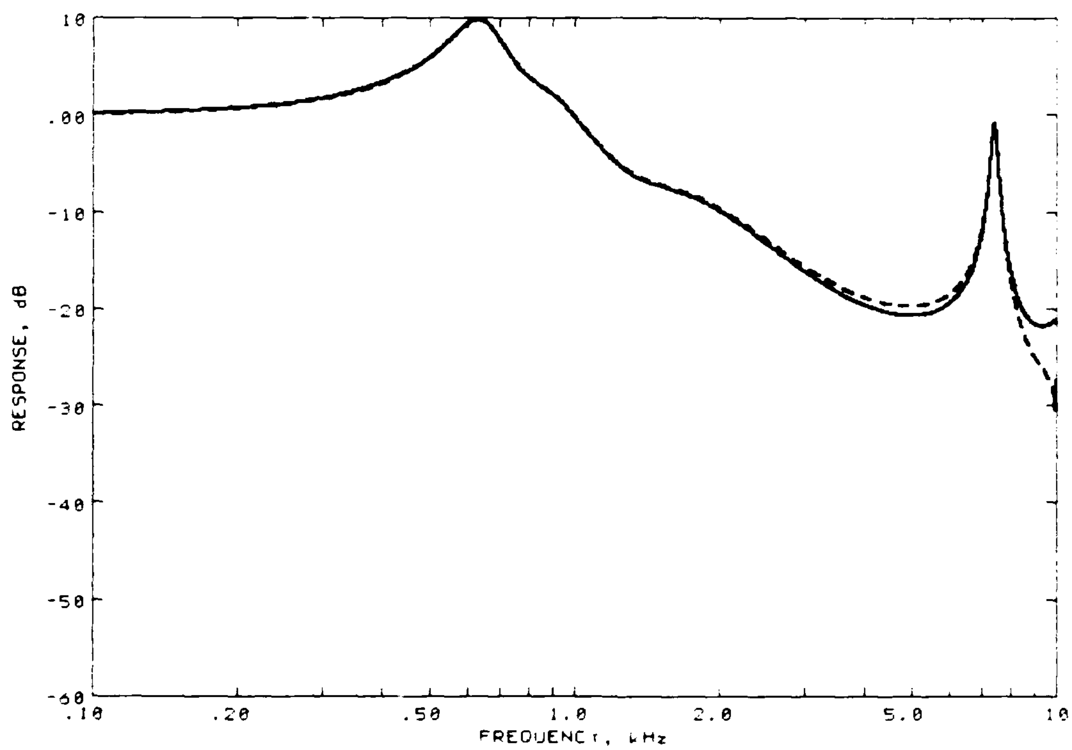

Figure 12.
Pressure transfer function for the direct signal path through the vent as specified in **Table 1**. The solid line is the target response, the dashed line the simulation.

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Table 2.
Parameters of the linear hearing aid without clipping.

Microphone:	Knowles EA-1842
Receiver:	Knowles ED-1913
Amplifier	
Type:	Current Source
Gain:	32 dB
Clipping level:	120 dB SPL
Crossover frequency:	2 kHz
Low-frequency channel	
Gain:	0 dB
Compression ratio:	1:1
Compression threshold:	Does not apply
Attack time:	Does not apply
Release time:	Does not apply
High-frequency channel	
Gain:	0 dB
Compression ratio:	1:1
Compression threshold:	Does not apply
Attack time:	Does not apply
Release time:	Does not apply

The major features of the curve are the two response peaks at about 2.2 and 5.4 kHz due to the receiver, and the low-frequency slope from the microphone.

The frequency response of the vented hearing aid is given by the solid line of **Figure 13**. The gain of the hearing aid amplifier is still 32 dB. At low frequencies, the resonance of the vent with the ear canal has resulted in a peak at about 600 Hz. Below the resonance peak, the gain is reduced due to the high-pass filtering provided by the vent frequencies below 200 Hz, the direct signal path through the vent starts to dominate the response and the slope of the curve is reduced. The gain at 100 Hz is therefore about 0 dB since most of the signal energy in this frequency region comes from the direct path.

The vent also has a strong effect at high frequencies. The feedback has increased the Q of both receiver resonances, resulting in peaks that are higher by several dB. The exact nature of the feedback effects, however, depends on the phase as well as the magnitude of the frequency response. Changing the phase response of the hearing aid processing will thus alter the details of the feedback effects in terms of which peak is more strongly affected and the maximum gain for which the system remains stable. An additional effect that is visible in the frequency response of **Figure 13** is the notch in the receiver output at 7 kHz due to the vent resonance.

Tone-burst response

In addition to modeling the behavior of a linear system with feedback, the time-domain simulation can also be used to model the behavior of a nonlinear system. Two common nonlinearities that occur in hearing aids are clipping distortion (due to amplifier saturation), and amplitude compression. The temporal response of a simulated hearing aid to a 1-kHz tone burst having a root mean square (RMS) level of 85 dB SPL was computed to illustrate the behavior of a hearing aid when nonlinear processing is present in the instrument.

The reference condition is the vented linear hearing aid specified in **Table 2** and used for the frequency-response of **Figure 13** (solid curve). The first 400 samples (20 ms) of the response of the linear instrument to the 1-kHz tone burst are shown in **Figure 14**. The amplitude scale is arbitrary, but is the same for all three figures in this section. The initial portion of the response shows the transients due to the sudden application of the sinusoid. There is a small amount of overshoot combined with components at other frequencies excited by the rising edge of the tone-burst. The envelope of the system response then oscillates slightly before reaching steady-state output at about 200 samples (10 ms).

The response to the same tone-burst was then repeated for the same hearing aid, but with amplifier clipping at an input-referred peak level of 85 SPL. The amplifier will thus clip the peaks of the input sinusoid. The hearing aid parameters are given in **Table 3**. This condition is still referred to in the industry as "linear" when discussing hearing aid processing, despite the presence of amplifier clipping or saturation, since no intentional nonlinear processing such as automatic gain control (AGC) has been designed into the instrument.

The tone-burst response is shown in **Figure 15**, where one can see that the clipping distortion has modified both the envelope of the transient response and the waveform shape of the steady-state response. The initial transient, especially between samples 20 (1 ms) and 80 (4 ms), shows a greater amount of high-frequency energy. The overshoot in this region, however, is similar to **Figure 14** since the clipping occurs before the receiver response and the transients, even when clipped, are still shaped by the receiver transfer function as modified by the vent and feedback. The steady-state response is at a slightly lower level than for the system without clipping. The distorted waveform shape clearly shows the presence of harmonic distortion. Thus, the effects of clipping are evident in the simulated temporal response.

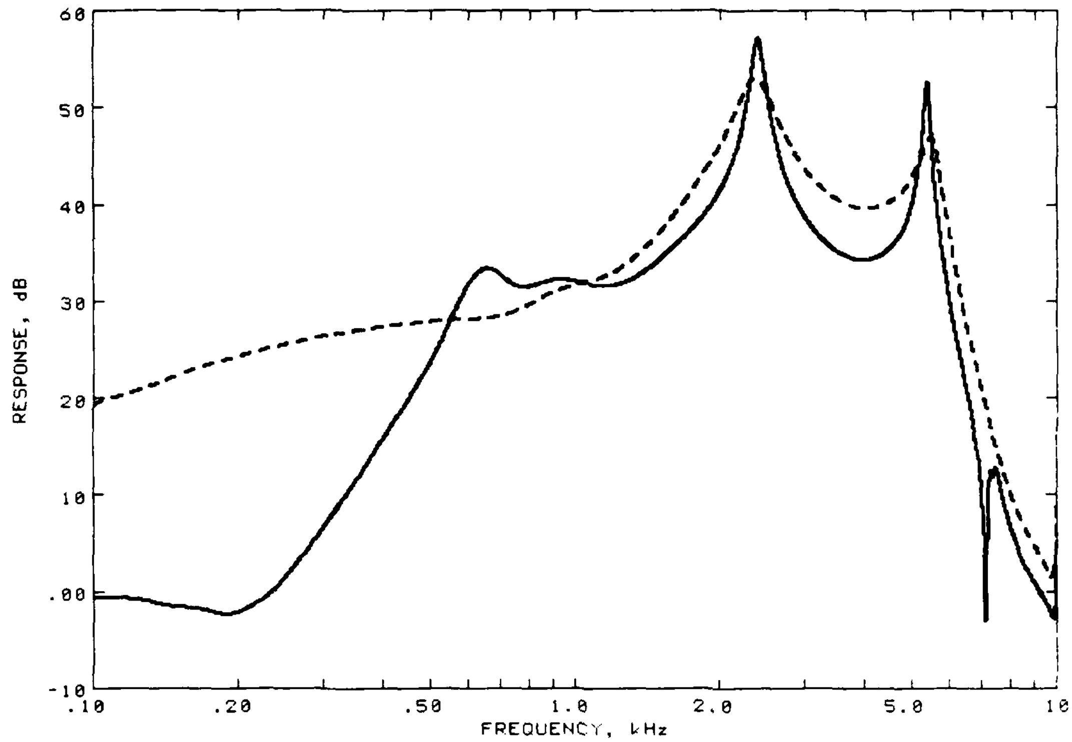

Figure 13.
Frequency response for the linear system with feedback as specified in **Table 2.** *The solid line is the response for the vent open.* the dashed line the response for the vent blocked.

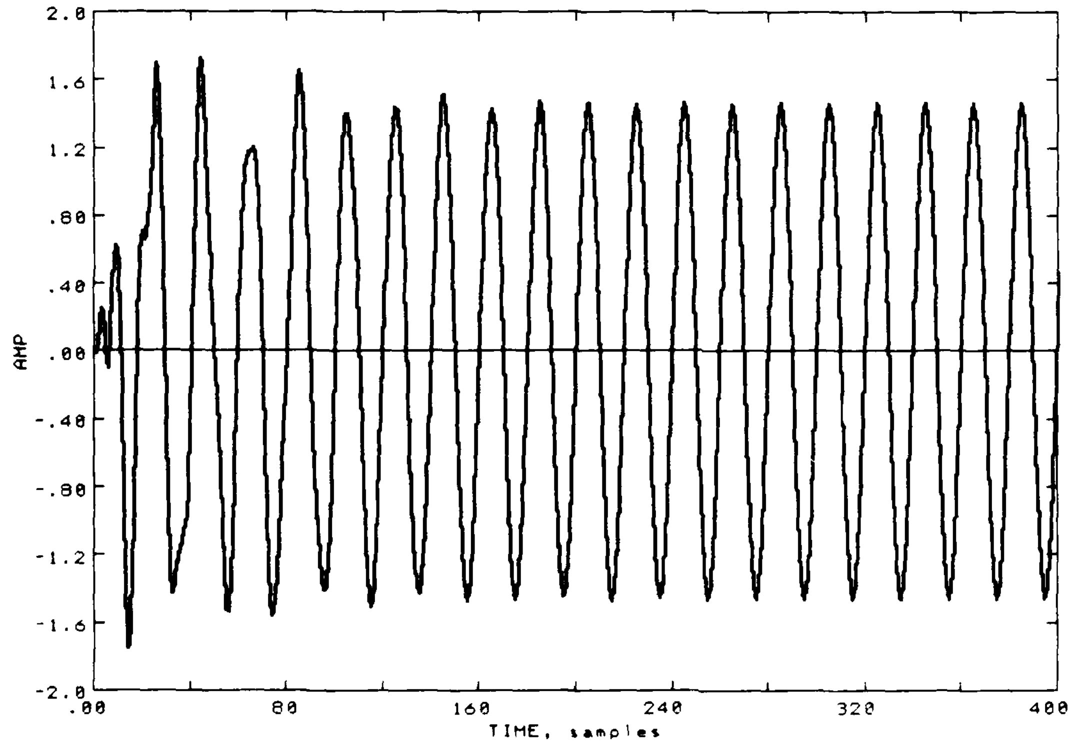

Figure 14.
Response of the simulated linear hearing aid of **Table 2** to an 85 dB SPL tone burst at 1 kHz. The sampling rate is 20 kHz.

Table 3.
Parameters of the linear hearing aid with clipping.

Microphone:	Knowles EA-1842
Receiver:	Knowles ED-1913
Amplifier	
Type:	Current Source
Gain:	32 dB
Clipping level:	85 dB SPL
Crossover frequency:	2 kHz
Low-frequency channel	
Gain:	0 dB
Compression ratio:	1:1
Compression threshold:	Does not apply
Attack time:	Does not apply
Release time:	Does not apply
High-frequency channel	
Gain:	0 dB
Compression ratio:	1:1
Compression threshold:	Does not apply
Attack time:	Does not apply
Release time:	Does not apply

Table 4.
Parameters of the two-channel AGC-I hearing aid

Microphone:	Knowles EA-1842
Receiver:	Knowles ED-1913
Amplifier	
Type:	Current Source
Gain:	32 dB
Clipping level:	85 dB SPL
Crossover frequency:	2 kHz
Low-frequency channel	
Gain:	0 dB
Compression ratio:	4:1
Compression threshold:	75 dB SPL
Attack time:	1 ms
Release time:	50 ms
High-frequency channel	
Gain:	0 dB
Compression ratio:	2:1
Compression threshold:	65 dB SPL
Attack time:	1 ms
Release time:	50 ms

The response of a two-channel compression hearing aid to the 85 dB SPL tone-burst is shown in **Figure 16**. The hearing aid parameters are given in **Table 4**; the gains in the two channels are the same but the compression characteristics are different. The frequency response of the compression hearing aid will be identical to that of the previous examples for an input below the compression thresholds, but will differ for an input such as the tone burst which is above threshold. The initial transient for the compression instrument is different than for the previous examples primarily because the fast compression attack has limited the amplitude of the 1 kHz tone. Thus, the transient components due to the receiver response are more readily visible for this case. The steady-state portion of the response shows the decreased amplitude of the tone due to the compression, with no apparent distortion for the pure-tone excitation.

Intermodulation distortion

A test signal consisting of two tones was used to illustrate the ability of the time-domain model to simulate intermodulation distortion effects. The test signal consisted of a pair of sinusoids at 1.0 and 1.2 kHz, each at an RMS level of 82 dB SPL. The resultant signal had an RMS level of 85 dB SPL and a crest factor of 6 dB. The simulation sampling rate was again 20 kHz. The output spectrum computed from 4,096 samples of the steady-state response of the vented linear hearing aid specified in **Table 2** is shown in **Figure 17**. The dB scale is arbitrary, but is the same for **Figure 17, Figure 18**, and **Figure 19**. The peaks due to the two sinusoids are quite apparent, and there is essentially no distortion. Introducing amplifier clipping at a level of 85 dB SPL, as specified in **Table 3**, results in the spectrum of **Figure 18**. There are a large number of distortion products generated by the clipping, with major peaks found in the vicinity of the third and fifth harmonics of the excitation sinusoids. The distortion spectrum is shaped by the receiver frequency response with its peaks at 2.2 and 5.4 kHz, since the receiver follows the amplifier with its distortion, so distortion products in the vicinity of the receiver peaks are given additional emphasis. Adding the compression specified in **Table 4** results in the output spectrum of **Figure 19**. The compression processing results in a small amount of intermodulation distortion, typically about 1 to 2 percent, but this level of distortion is substantially lower than the results of amplifier saturation for the same input level and clipping threshold.

CONCLUSIONS

This paper has presented a time-domain digital computer simulation of an ITE hearing aid. The simulation includes a microphone, two-channel compression process-

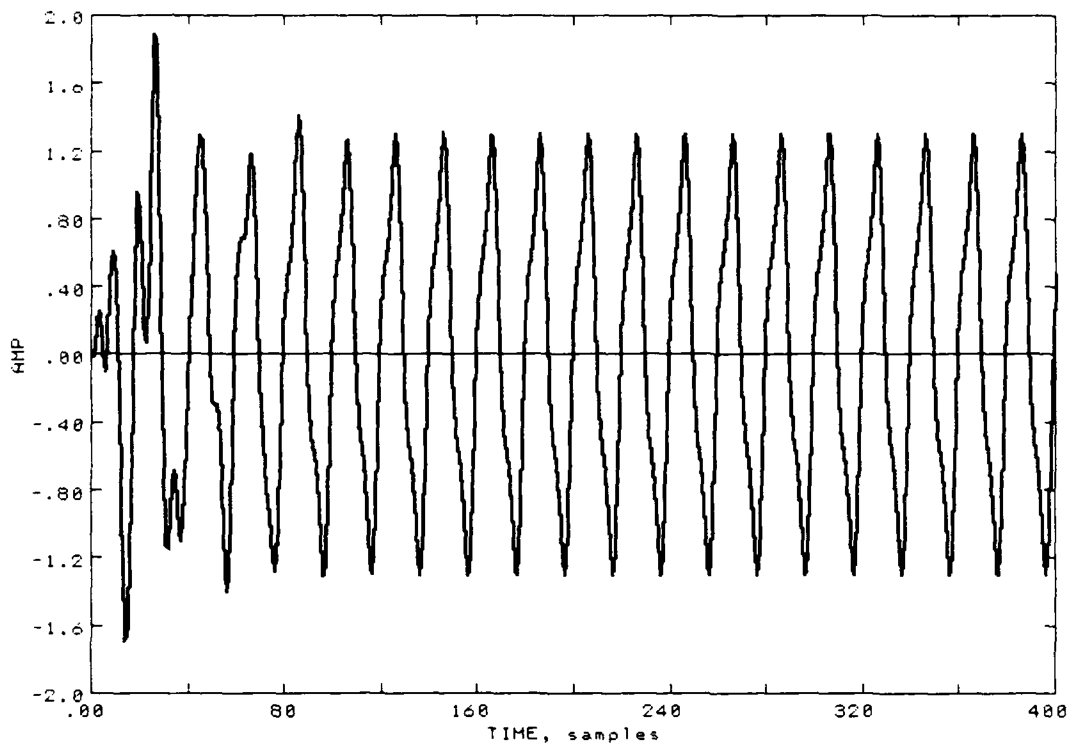

Figure 15.
Response of the simulated linear hearing aid with clipping, as specified in **Table 3**, to an 85 dB SPL tone burst at 1 kHz. The sampling rate is 20 kHz.

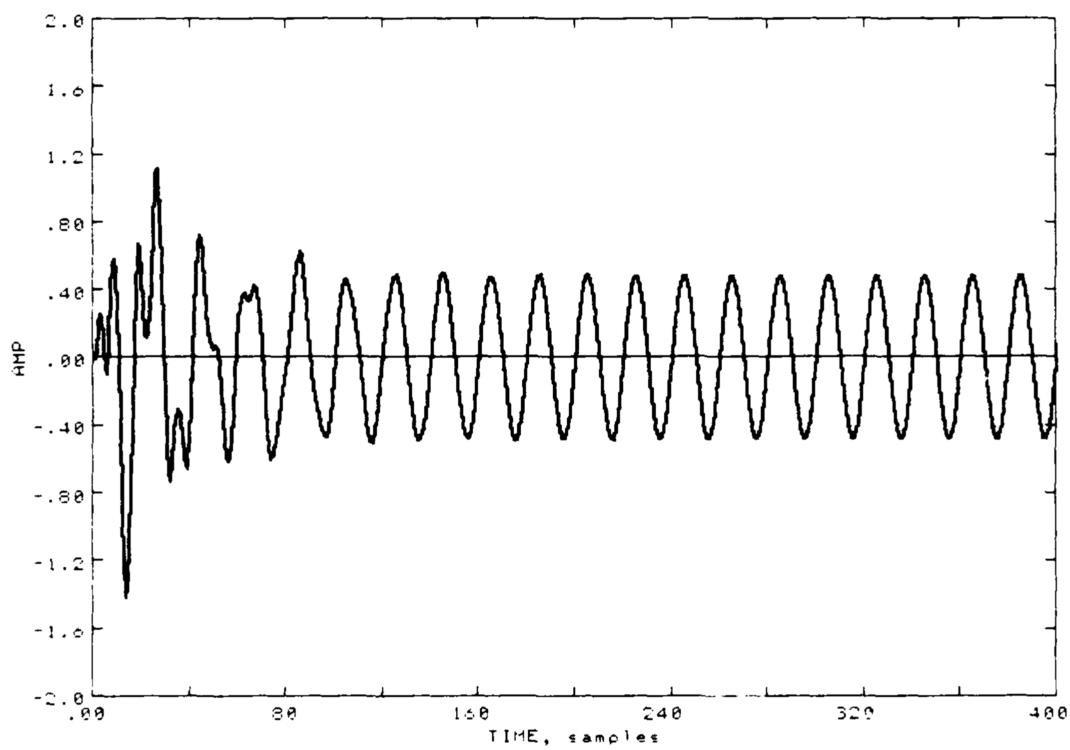

Figure 16.
Response of the simulated two-channel compression hearing aid of **Table 4** to an 85 dB SPL tone burst at 1 kHz. The sampling rate is 20 kHz.

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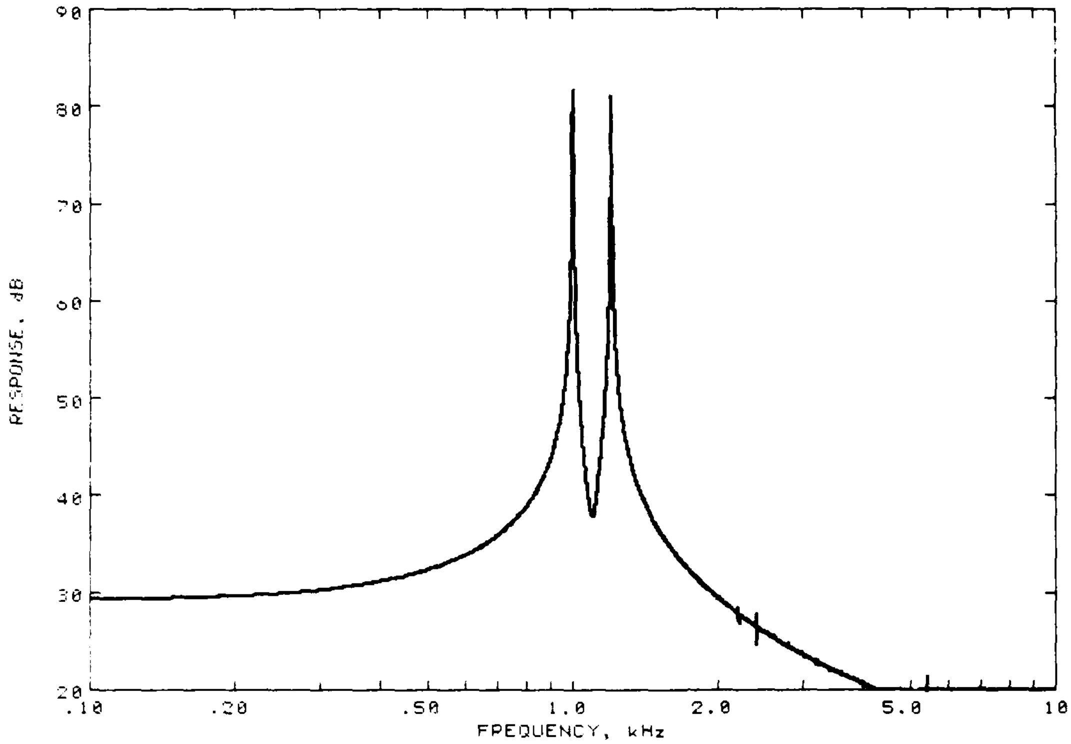

Figure 17.
Response of the simulated linear hearing aid of **Table 2** to a two-tone excitation at 85 dB SPL. The sampling rate is 20 kHz.

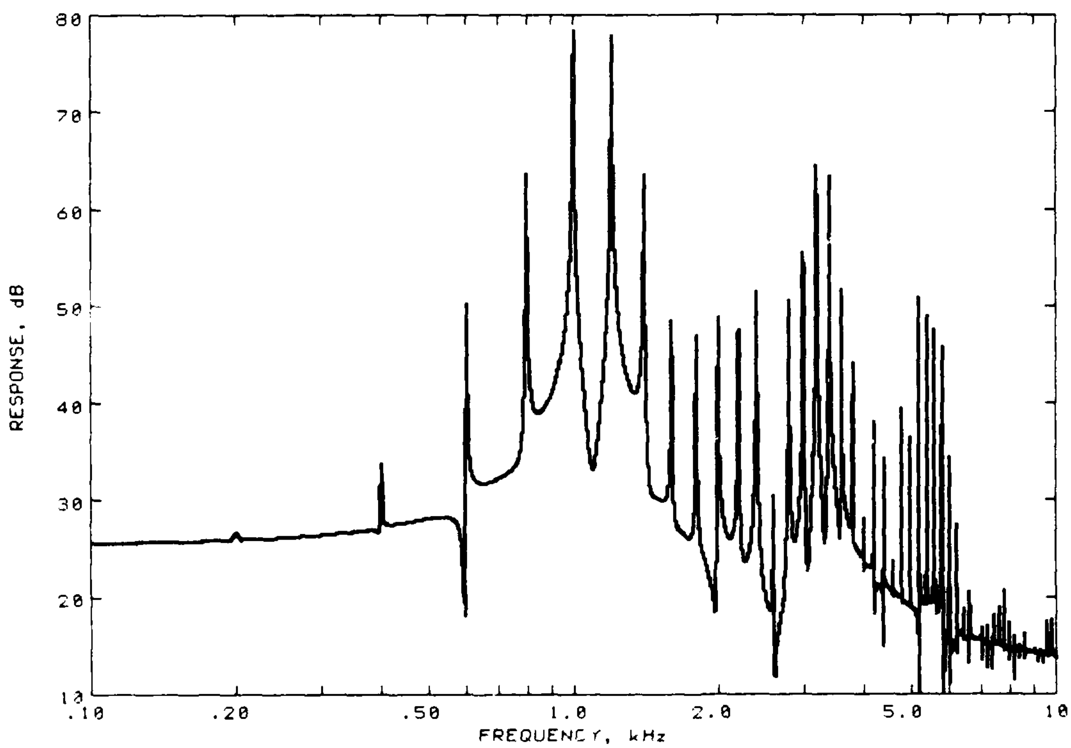

Figure 18.
Response of the simulated hearing aid with clipping, as specified in **Table 3**, to the two-tone excitation at 85 dB SPL. The sampling rate is 20 kHz.

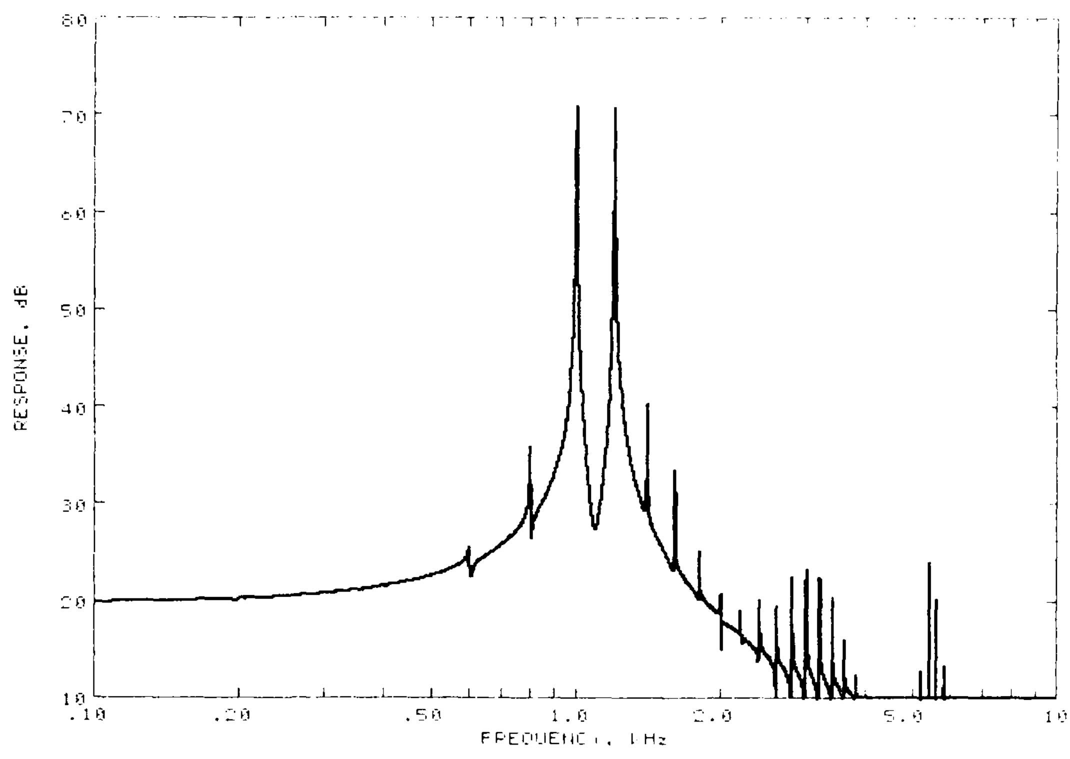

Figure 19.
Response of the simulated two-channel compression hearing aid of **Table 4** to the two-tone excitation at 85 dB SPL. The sampling rate is 20 kHz.

ing, an amplifier with clipping distortion, a receiver, an ear canal and ear drum, and a vent. The model allows for the simulation of acoustic feedback, and goes beyond existing frequency-domain models in also simulating nonlinear effects such as AGC-I and amplifier saturation. The model has been implemented using floating-point arithmetic, so quantization effects in the digital filters are not significant.

Because it is a digital transformation of an analog system, the simulation does not provide an exact reproduction of the analog hearing aid. This is apparent in the slight deviations in portions of the frequency response curves presented for some of the elements of the simulation. Greater accuracy can be obtained by using longer digital filters and a higher sampling rate, but this increases the computational burden without providing a corresponding increase in the amount of information produced. The simulation is thus a representation of a general class of hearing aids rather than a reproduction of a specific instrument.

The purpose of the time-domain simulation is to illustrate aspects of hearing aid behavior and to serve as a computer test bed for the development of new hearing aid processing algorithms and test approaches. The basic simulation can be modified to include more processing channels, different forms of compression, speech enhancement, adaptive filters for feedback suppression, or any other processing scheme to be developed or evaluated. Different portions of the simulation, such as the vent or a specific processing option, can be adjusted or turned on or off at will to give a convenient way of studying the effects of modifying the instrument or the acoustic environment. Because the simulation is equivalent to a self-contained, computer-controlled hearing aid, it can also be used to process stimuli for the development of test signals and processing techniques for the characterization of existing or future hearing aids. The time-domain simulation is therefore a powerful tool for the study of hearing aid behavior.

ACKNOWLEDGMENT

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A Review Article

Wheelchair racing sports science: A review

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Abstract—Wheelchair racing science and the performance of athletes involved in wheelchair racing have developed rapidly in recent years. With increasing interest in this sport, the need arises to identify areas where further research is necessary and cooperation between individuals with various backgrounds is encouraged. Many of the problems facing investigators in this field require knowledge in several areas of science and engineering, which suggests an interdisciplinary approach to these issues. Further progress would also benefit from the development of more quantitative methods for the classification of wheelchair athletes, or a restructuring of the classification system; development of sophisticated instrumentation for racing wheelchairs; standardization of test procedures and more complete reporting of results of studies; and, more in-depth mathematical modeling and computer simulation of wheelchair racing. This review presents an overview of four areas of wheelchair racing science: 1) classification of wheelchair athletes; 2) design and analysis of racing wheelchairs; 3) biomechanics of racing wheelchair propulsion; and, 4) training and coaching of wheelchair racers.

Key words: *athletic training, biomechanics, classification of wheelchair athletes, instrumentation, racing wheelchair, sports psychology.*

INTRODUCTION

There are no statistics kept on the number of participants involved in wheelchair sports. However, racing wheelchair manufacturers estimate that over the last 5 years more than 10,000 racing wheelchairs have been manufactured worldwide, with the majority being produced in Europe, Japan, and North America. The number of participants in wheelchair sports, especially wheelchair racing, is growing rapidly, as is apparent in the number of wheelchair racing participants competing in major road races (53).

The growing number of competitors and the increasing quality of the competition have fostered interest in wheelchair sports science, with wheelchair racing and basketball leading the field (10). **Figure 1** is a typical example of wheelchair racing.

There are numerous studies of persons with spinal cord injuries as applied to sports and exercise (25,26). This review represents a significant portion of the work published in the field of wheelchair racing science.

HISTORY AND DEVELOPMENT OF WHEELCHAIR RACING

Shortly after World War II, Sir Ludwig Guttman and his colleagues originated wheelchair sports as a rehabilitation tool at Stoke Mandeville Hospital in England (63). This developed out of the need to provide exercise and recreational outlets for the large number of young persons recently injured in the war. News of Dr. Guttman's success with the rehabilitation of his patients through the use of sports soon spread throughout Europe and to the United States. In 1948, he organized "Games" for disabled British veterans. In 1952, the Games developed into the first international wheelchair sporting competition for the

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Figure 1.
An example of wheelchair racing. (The final of the Class V 1500-meter at the 8th Paralympic Games.)

disabled, with participants from the Netherlands, the Federal Republic of Germany, Sweden, and Norway. During this event, the International Stoke Mandeville Games Federation (ISMGF) was formed to govern and develop wheelchair sporting competitions; an organization that continues to be the international governing body for wheelchair sports. The ISMGF established ties to the International Olympic Committee (IOC), thus expanding the scope of wheelchair sports.

The first international games for the disabled held in conjunction with the Olympic Games took place in 1960 in Rome, Italy. The name "Paralympics" was coined during the 1964 Tokyo games and, as such, subsequently held every four years.

In the early years of wheelchair racing, participants used bulky standard wheelchairs and did not compete in events with distances over 200 meters. In the 1970s, athletes started to modify their wheelchairs for specific sports and began to take an interest in road racing.

In 1975, a young paraplegic became the first person to compete in the Boston Athletic Association Marathon in a wheelchair. This opened the door for many future wheelchair road racers, prompting Dr. Caibre McCann, a leading physician for the ISMGF international governing body for wheelchair sports, to say: "Running is natural, but propelling yourself in a wheelchair is an unnatural phenomenon. People never realize what a wheelchair athlete is capable of. This is a breakthrough in man's limits."[1] Within a few years, several nationally recognized road races initiated wheelchair divisions and more disabled persons began to train for these races than had ever been anticipated. In 1976, the ISMGF started to coordinate with other international sports organizations to launch a unified international disabled sports movement (46,50).

Racing wheelchairs began to evolve as special-purpose pieces of equipment easily distinguishable from everyday wheelchairs. Distances on the track were extended to include races up to 1500 meters, and during this transition, the mile record was dropped to below 5 minutes.

The early 1980s saw the development of more sophisticated racing wheelchairs and training techniques. By 1985, most racing wheelchairs no longer had any components in common with everyday wheelchairs (which had also improved dramatically), and George Murray became the first wheelchair racer to break the 4-minute mile. In the years that followed, wheelchair racing continued to progress with improved equipment, training, and nutrition; consequently, world records were continuously being broken. Wheelchair racing began the path toward recognition as a legitimate Olympic sport in 1984 when the men's 1500 meter and the women's 800 meter wheelchair races were included as demonstration events in the Olympic Games held in Los Angeles, CA.

In 1988, the 8th Seoul Paralympics were held with over 60 countries represented by approximately 4,000 athletes—the largest games to date. In addition, the 24th Olympiad included the men's 1500 meter and women's 800 meter wheelchair races as demonstration events. The record books were rewritten and many technical advances were apparent.

THE WHEELCHAIR ATHLETE

The two major components of wheelchair racing are the chair and the athlete. This section will give a brief description of the literature available on wheelchair athletes. The number of subjects used in studies of these athletes is quite small: the maximum sample size for metabolic data is 15, most are less than ten. The subjects are generally grouped by their level of impairment due to disability (quadriplegia and paraplegia). Practically all work in the available literature has concentrated on males; qualified female subjects are apparently difficult to find.

Physiological testing

Table 1 shows the metabolic results for several studies of wheelchair athletes (24,28,52). There is a fair amount of variation between the studies, as might be expected for the small sample sizes. The average paraplegic values for all of the studies are 185.3 bpm and 2.14 l/min which are similar in magnitude to able-bodied subjects doing upper body exercise. However, the values for the quadriplegic subjects are noticeably lower (121.5 bpm and 0.74 l/min); this is because they do not have the functional muscle mass to elicit a higher response and their central and autonomic nervous systems are modified to a greater extent.

Only two studies of postexercise blood lactate levels are available. Pitetti, Snell, and Stray-Gundersen (59) found the 3-minute post-maximal exercise blood lactate levels to be 8.1 ± 0.7 mmol/l, while Pohlman, Gayle, Davis, and Glaser (60) found the difference between pre- and post-maximal exercise blood lactate levels to be between 2.3 and 2.6 mmol/l. No report was made on the time samples were taken.

Table 1.
Metabolic data for wheelchair athletes.

Source	Max heart rate (beats/min)	VO_2 max (l/min)
	Paraplegics	
Davis & Shepard (1988) (n=15)	181.7 ±9	2.24±0.14
Pitetti, Snell, & Gundersen (1987) (n=8)	180 ±2	1.90±0.1
Lakomy, Cambell, & Williams (1987) (n=10)	193 ±15	1.95±0.38
Coutts & Stogryn (1987) (n=4)	190.25±9.65	2.74±0.78
	Quadriplegics	
Figoni, Boileau, Massey & Larsen (1988) (n=11)	122 ±8	0.66±0.07
Lakomy, Cambell, & Williams (1987) (n=2)	119 ±8.5	1.15±0.07

Table 2.
1988 NWAA national track records.

Meters	Men		Women	
	Time in seconds			
	Quad	Para	Quad	Para
100	20.1	16.8	23.1	17.7
200	40.1	32.2	54.4	35.4
400	79.1	57.5	90.1	67.9
800	164.8	120.4	184.4	138.1
1500	308.9	225.8	356.0	270.9
3000	689.1	N A	****	N A
5000	****	777.3	****	902.2

Track records

Table 2 lists the 1988 National Wheelchair Athletic Association (NWAA) records for track racing (condensed to open quadriplegic and paraplegic divisions for men and women). As expected, the records for the quadriplegics are slower than those of the paraplegics. The same is true of women as compared to men.

Figure 2 presents a plot of average speed for each national record against distance (time/distance versus distance). The maximum average speed for all of the divisions occurs in the 400-meter race; perhaps a result of the longer time needed to reach maximum speed.

Although no official national records (national bests) are kept for road racing, **Table 3** shows the analysis of 200 wheelchair road racing results. As expected, the marathon speeds are slower than the 10,000-meter speeds; however, both speeds are much greater than those of runners. In addition, elite wheelchair athletes are capable of attaining peak speeds in excess of 8.5 meters per second (19 mph) on level ground, and 17.9 meters per second (40 mph) on downgrades.

Studies needed

Much has been done in this area, but the work is often unrelated and difficult to apply to wheelchair racing. Studies need to be performed using a number of different protocols (i.e., increasing speed, increasing resistance, increasing speed and resistance) on different devices (i.e.,

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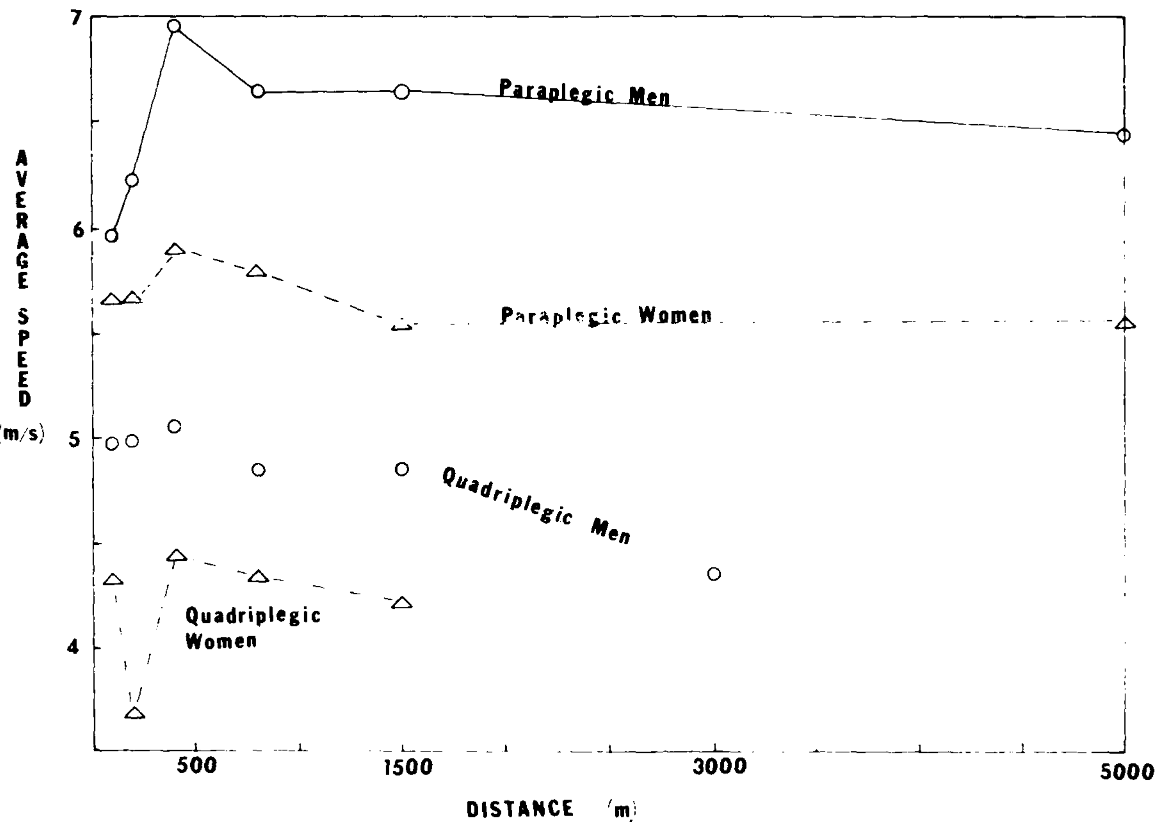

Figure 2.
Plot of average speed versus distance based upon the 1988 National Wheelchair Athletic Association (NWAA) records.

wheelchair ergometers, arm-crank ergometers, and treadmills). From these tests, we may be able to relate the results of different investigators to our own work. More emphasis should be placed on testing actual wheelchair users whenever possible. Studies need to be performed during actual or simulated race and training situations, so that we can relate laboratory results to practice. Relationships need to be developed to relate metabolic, cardiorespiratory, and anthropometric values to training and racing. The biodynamics of interacting with other competitors needs to be studied (e.g., the energy savings while drafting). New portable instruments need to be developed to simplify the collection and recording of mechanical, metabolic, and cardiorespiratory data during actual training and competition.

Table 3.
Long distance road racing results.

	First 10 to finish			First to finish		
	Marathon	10,000m	400m	Marathon	10,000m	400m
Time (s)	7340.5 ± 605.8	1576.4 ± 103.1	57.5*	6663.1 ± 312.5	1501.1 ± 66.2	***
Speed (m/s)	5.83 ± 0.46 (13.0 mph)	6.37 ± 0.41 (14.2 mph)	6.96 (15.6 mph)	***	***	***
Shortest time (s)	6205	1380		***	***	***
FAS** (m/s)	6.85 (15.3 mph)	7.25 (15.2 mph)		***	***	***

*National record
**Fastest average speed

Table 4.
Common characteristics of wheelchairs.

	Racing		Standard	
	Meters	Inches	Meters	Inches
Front wheel	0.36-0.46	14-18	0.13-0.20	5.1-7.9
Rear wheel	0.61-0.69	24-27	0.56-0.66	22.0-30.0
Push-ring	0.25-0.46	12-18	0.46-0.56	18.1-22.0
Push-ring tubing	0.0095-0.0300	3/8-1.2	0.0159-0.0381	3.5-1.5
Overall length	1.20-1.50	47.25-59.1	0.91-1.14	35.8-44.9
Wheelbase	0.60-1.05	23.5-41.3	0.36-0.61	14.2-24.0
	Kilograms	Pounds	Kilograms	Pounds
Frame weight	1.35-3.2	3-7	2.7-10.0	6.0-22.1
Chair weight	4.5-7.3	10-16	6.5-18.0	14.3-39.7

DESIGN AND ANALYSIS OF RACING WHEELCHAIRS

Little has been reported about racing wheelchairs in the technical and scientific literature. These chairs are specialized sole-purpose pieces of equipment with distinguishing features (i.e., tubular tires, lightweight rims, precision hubs, larger wheels, and smaller push-rings, etc.) designed to optimize an individual's racing ability. **Table 4** presents an overview of some of the common characteristics of racing wheelchairs. **Figure 3** shows a drawing of a racing wheelchair with the major parts labeled.

At this time, the tools of engineering and mathematics have been applied to a very limited degree (11,12,48). The use of sophisticated analyses available from signal processing, system identification, and control theory, when used in conjunction with the knowledge to be gained from physiological, medical, biomechanical, and psychological tools, will increase the present understanding of the effects of wheelchair propulsion; specifically, the effect of wheelchair racing on the mobility-impaired individual.

Two recent papers (15,19) describe the features of racing wheelchairs in detail. With the present state of wheelchair racing, describing such a chair is like hitting a rapidly

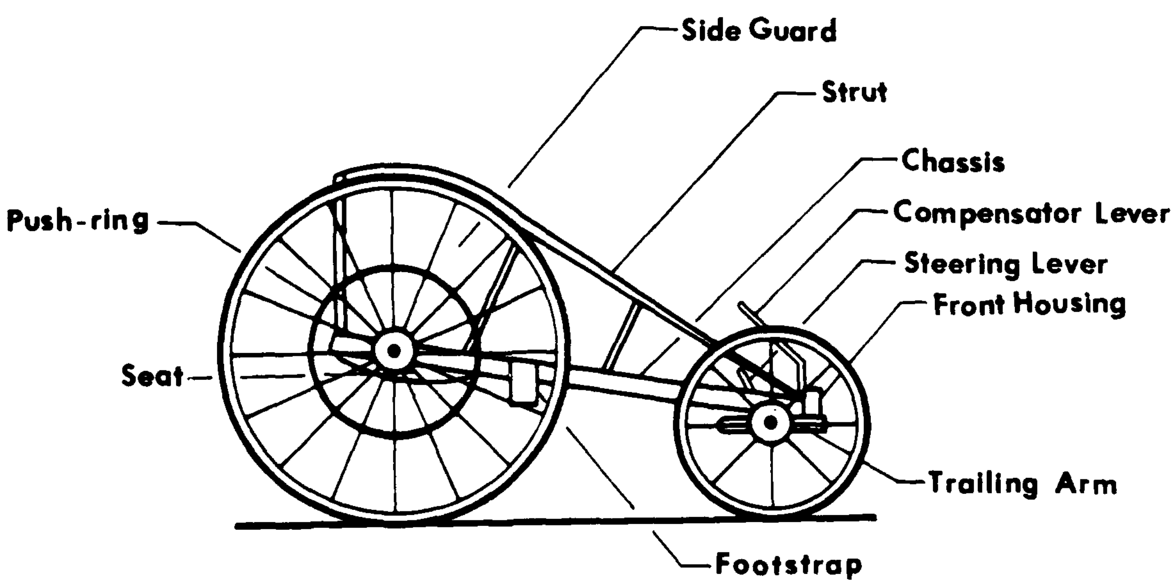

Figure 3.
A typical racing wheelchair.

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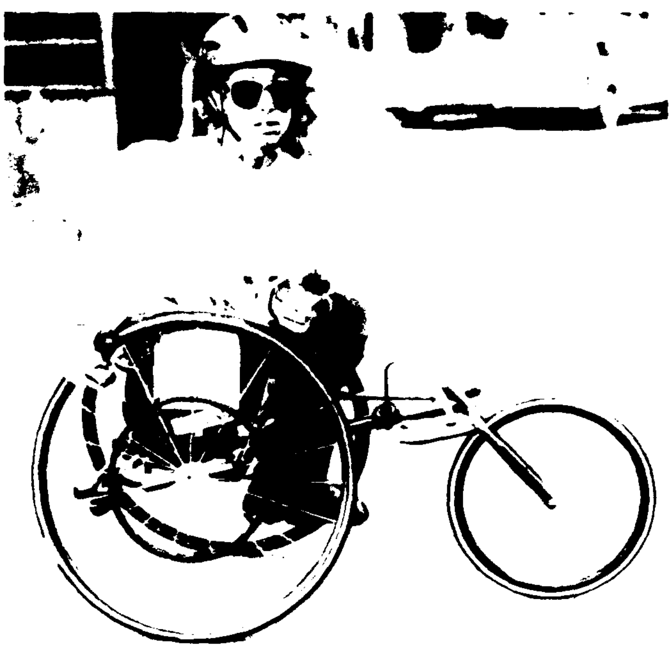

Figure 4a.
Side view of an athlete in a racing wheelchair.

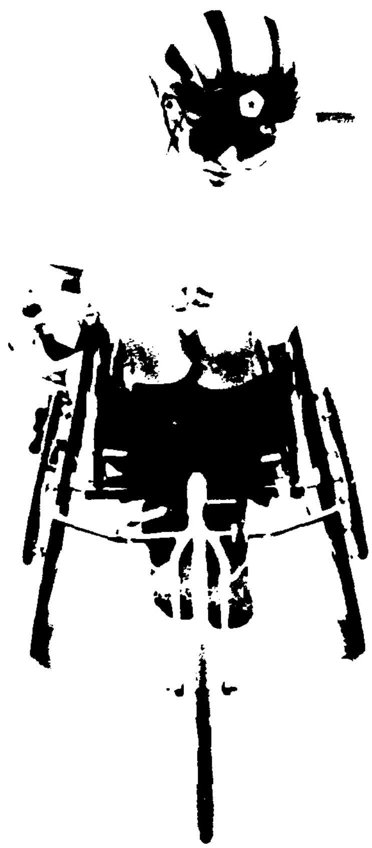

Figure 4b.
Front view of an athlete in a racing wheelchair.

moving target. For example, two years ago, three-wheeled racing wheelchairs were an uncommon sight; today they represent a significant portion of the racing wheelchair market. **Figures 4a** and **4b** show an athlete in a typical racing wheelchair, and the relationship between the two. Notice the tight fit of the chair to the athlete's body.

Many factors affect the efficiency of racing wheelchairs: weight and balance; frame and wheel stiffness; rolling, bearing, and air resistance; and the frame geometry. Most of these factors have not been studied as they relate to racing wheelchairs, and none has been completely defined.

Free body diagrams for the motion of a racing wheelchair on flat ground (no road crown or road irregularities) and on an inertial dynamometer are given in **Figures 5a** and **5b**. The necessary conditions for equivalence of the corresponding differential equations for the motion have been derived by Cooper (17). Various dynamometers have been described in the literature (29,70). These dynamometers are based upon the use of electric motors for dynamic loading. The California State University at Sacramento has, perhaps, the most sophisticated wheelchair dynamometer: it is computer-controlled and capable of simulating a number of different course scenarios and control algorithms (17). Their dynamometer simultaneously measures and controls wheel torque, speed, and power. They are presently working on controlling heart rate and developing some interactive video games.

York and Kimura (78), as well as Higgs (42), have conducted investigations on basic construction variables without conclusive results. These studies focused on determining the differences and similarities across class, and the differences and similarities of sprinters and distance racers. Few differences were found because: 1) many athletes compete in both sprints and distance races; and, 2) racing wheelchairs are highly individualized.

Cooper (19) described the basic principles behind racing wheelchair design, but did not present a detailed analysis. In general, there is the classical trade-off between weight and rigidity when designing a racing wheelchair. Because the available power of the athlete is small, energy loss between the athlete and the ground must be minimized (but with a restriction on the additional weight of the chair). Racing wheelchairs weigh between 4.5 and 7.3 Kg, with the center of gravity nominally located along the center line. The most commonly used materials for frame construction are *Chromolly* steel (SAE 4130) and aluminum (SAE 6061). These materials are readily available, have good strength-to-weight ratios, and are easy to work with.

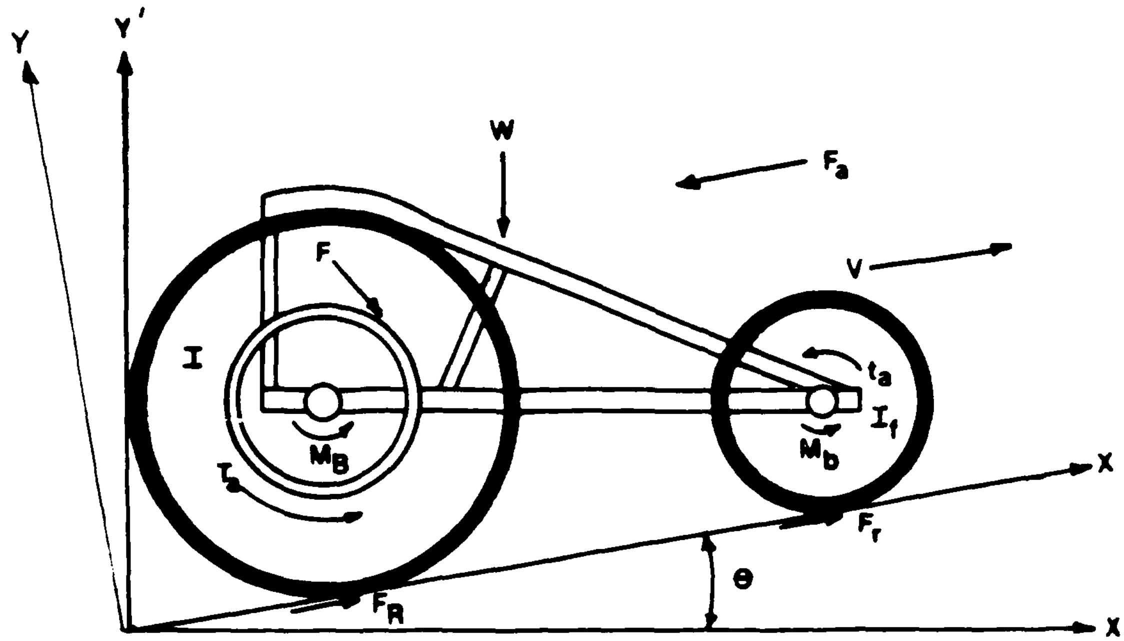

Figure 5a.
Free-body diagram for the motion of a racing wheelchair on a flat road (no road crown or irregularities).

The use of advanced composites has been investigated to a limited extent, primarily by Lawrence Livermore National Laboratories, the University of Delaware, and California Polytechnic State University at San Luis Obispo, but nothing has yet been published in this area. Golumbek[2] performed some preliminary investigations of several design factors for racing wheelchairs: finite element analysis of rear axles, and the design of a carbon fiber frame. He found that a 17-mm axle would provide a significant reduction in the axle-bending moments. Cooper has investigated the problem of rear wheel alignment (18), crown compensation (directional stabilization due to road crown) (16), and the use of frame geometry in directional control (14).

Researchers at the University of Virginia have investi-

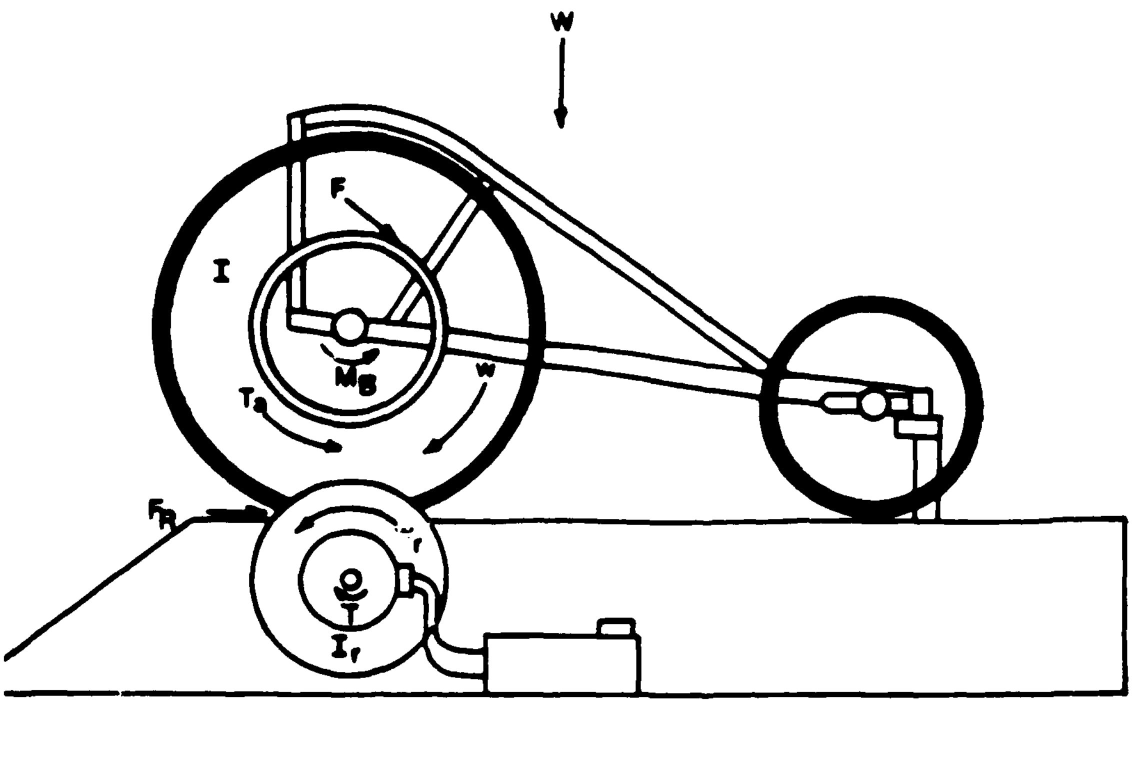

Figure 5b.
Free-body diagram for the motion of a racing wheelchair on an inertial dynamometer.

gated the various functional aspects of standard wheelchairs (5,6,7,11,47,48,49). **Figure 6** shows a photograph of a standard wheelchair. Although the results of their studies are not directly applicable, their methods form a basis for possible further investigation of racing wheelchairs.

Papers have been written on the stability of standard wheelchairs (11,76), but the stability of racing wheelchairs has yet to be investigated. It was found that front caster wheelchairs are directionally more stable than rear caster wheelchairs. This information may be useful in determining the solution to two problems facing wheelchair racers: At what speed does the racing wheelchair become laterally unstable? (**Figure 7**); and, What are the acceptable turning radii for various speeds in order to maintain roll stability? (**Figure 8**). Typically, when a vehicle becomes laterally unstable, it will rapidly veer off course if there is no steering. If someone is steering, the vehicle (if it has exceeded the stability limit) will oscillate back and forth about the desired heading until it goes off the road or track. This happens because the pilot tries to correct for the change in heading, and oversteers. Lateral instability primarily depends upon visibility, reaction time, task complexity, and the vehicle dynamics. Roll stability is also of particular importance. Racing wheelchairs (and their users) have been observed to roll over (flip) while turning, especially while going down a hill. Without a human pilot, the wheelchair will fall onto its side when speed is too great for the turn.

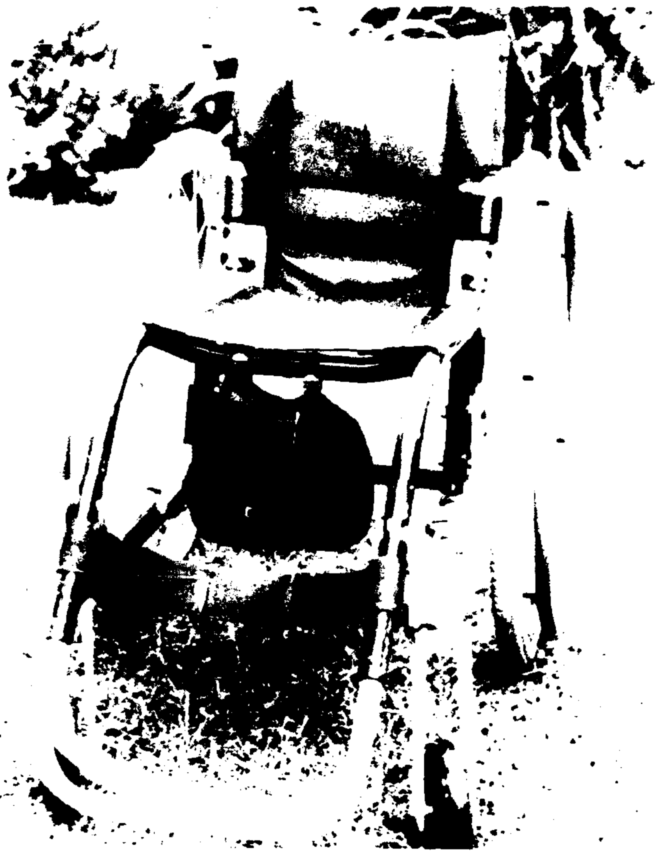

Figure 6.
Photograph of a standard sports wheelchair.

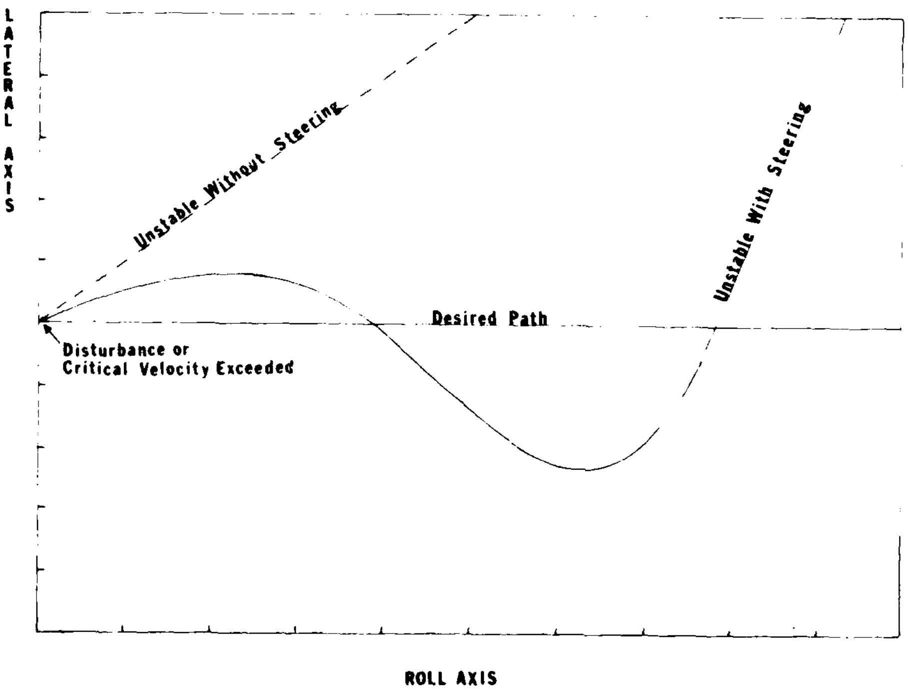

Figure 7.
A graph showing potential lateral instability curves for an impulse type disturbance (a wind gust acting upon the side of the individual/wheelchair). —— Instability curve assuming steering. - - - Instability curve without steering.

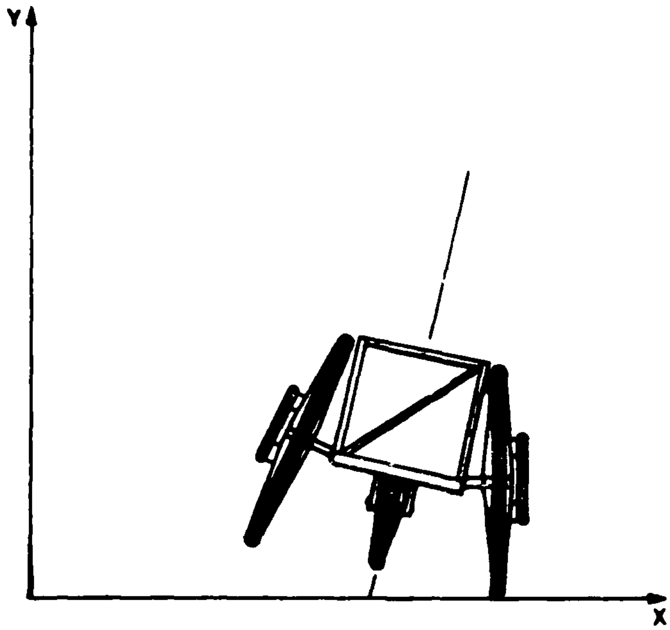

Figure 8.
Roll instability diagram. (Roll instability is described by the dynamics that determine when a chair will roll over or flip.) x = pitch axis; y = yaw axis.

With a human pilot, the wheelchair may bounce from balancing on one side of the chair to the other (because of the steering, as in the lateral stability case). If the individual is able to control the chair, the oscillations will dampen out (the chair may not be heading in the desired direction but will be balanced on all of its wheels); if not, the wheelchair/individual system will fall on its side. The relative stability of three- and four-wheel configurations is of particular interest.

Racing wheelchairs are highly individual. The factors determining the selection of the proper chair for an individual are not completely understood. In most cases, experience is the most appropriate tool.

More in-depth analyses of racing wheelchairs need to be performed to determine: 1) the location and magnitude of the stresses and forces acting upon the frames and the components; 2) the optimum steering geometry; and, 3) the magnitude of aerodynamic losses. These analyses should establish where improvements can be made and, where possible, failures can be prevented. Studies need to be performed to determine product life span and maintenance requirements. The tools of Finite Element Analysis need to be exploited to a greater extent. Dynamic stability analyses of racing wheelchairs need to be performed, evaluating various racing wheelchair configurations (four-wheeled chairs versus three-wheeled chairs) under a wide range of circumstances (down hills, turning, road surface irregularities, and interaction with other competitors). These studies must eventually account for the user control.

CLASSIFICATION OF WHEELCHAIR ATHLETES

Figure 9 presents a graphic explanation of the classification system used by the NWAA. There are seven classes (3 quadriplegic and 4 paraplegic) and open (classless) competition. Each of the seven classes is designed to represent an ascending level of physical ability.

Wheelchair athletes are classified by functional potential in order to encourage fair competition among individuals with similar levels of mobility impairment as related to manual wheelchair propulsion. Classification has been a source of controversy since its inception, because if an athlete is misclassified he/she may have an unfair advantage over fellow competitors. Problems arose in the mid-1950s when people began to notice that a single class for all wheelchair users was not equitable (quadriplegics never won any races in which paraplegics were involved). The number of classes has changed several times. There has been considerable discussion, but little scientific research concerning: 1) the basis upon which a classification system should be founded; and, 2) what constitutes a significant difference in potential needed to propel a racing wheelchair (20,25). Most of the reported research is concerned with methods of evaluating athletes under the present classification system. Weis and Curtis (77) and McCann (55,56) have written reviews on the controversies surrounding the classification of wheelchair athletes.

At present, the problems of classification are focused on the reduction in the number of classes and on the incorporation of athletes from multiple-disability groups. Organizers of large international events profess that the number of classes must be reduced in order to make wheelchair racing more understandable to the public (thus generating greater interest in it as a spectator sport), and because the logistics of organizing "so many" races is prohibitive. Others claim that competition becomes meaningless when there are so many classes, and no one wishes to see athletes discouraged or eliminated from competition by an unjust classification system.

Recently, the focus has been on various ways of developing a functional classification system in which athletes are classified by their potential ability to compete in the sport for which they are being classified. In contrast, the present system focuses on classifying people into "similar" degrees of disability. Functional classification emphasizes each athlete's physiology, whereas the present system emphasizes each individual's anatomy. The contro-

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CLASS IA

All cervical lesions with complete or incomplete quadriplegia who have involvement of both hands, weakness of tricepts (up to and including grade 3 on testing scale) and with severe weakness of the trunk and lower extremities interfering significantly with trunk balance and the ability to walk.

CLASS IB

All cervical lesions with complete or incomplete quadriplegia who have Involvement of upper extremities but less than 1A with preservation of normal or good tricepts (4 or 5 on testing scale) and normal or good finger flexion and extension (grasp and release) but without intrinsic hand function and with a generalized weakness of the trunk and lower extremities interfering significantly with trunk balance and the ability to walk.

CLASS IC

All cervical lesions with complete or incomplete quadriplegia who have involvement of upper extremities but less than 1A with preservation of normal or good tricepts (4 or 5 on testing scale) and normal or good finger flexion and extension (grasp and release) but without intrinsic hand function and with a generalized weakness of the trunk and lower extremities interfering significantly with trunk balance and the ability to walk.

CLASS II

Complete or incomplete paraplegia below T1 down to and including T5 or comparable disability with total abdominal paralysis or poor abdominal muscle strength (0-2 on testing scale) and no useful trunk sitting balance.

CLASS III

Complete or incomplete paraplegia or comparable disability below T5 down to and including T10 with upper abdominal and spinal extensor musculature sufficient to provide some element of trunk sitting balance but not normal.

CLASS IV

Complete or incomplete paraplegia or comparable disability below T10 to and including L2 without quadricepts or very weak quadricepts with a value up to and including 2 on the testing scale and gluteal paralysis.

CLASS V

Complete or incomplete paraplegia or comparable disability below L2 with quadriceps in grades 3-5.

Medical Classifications

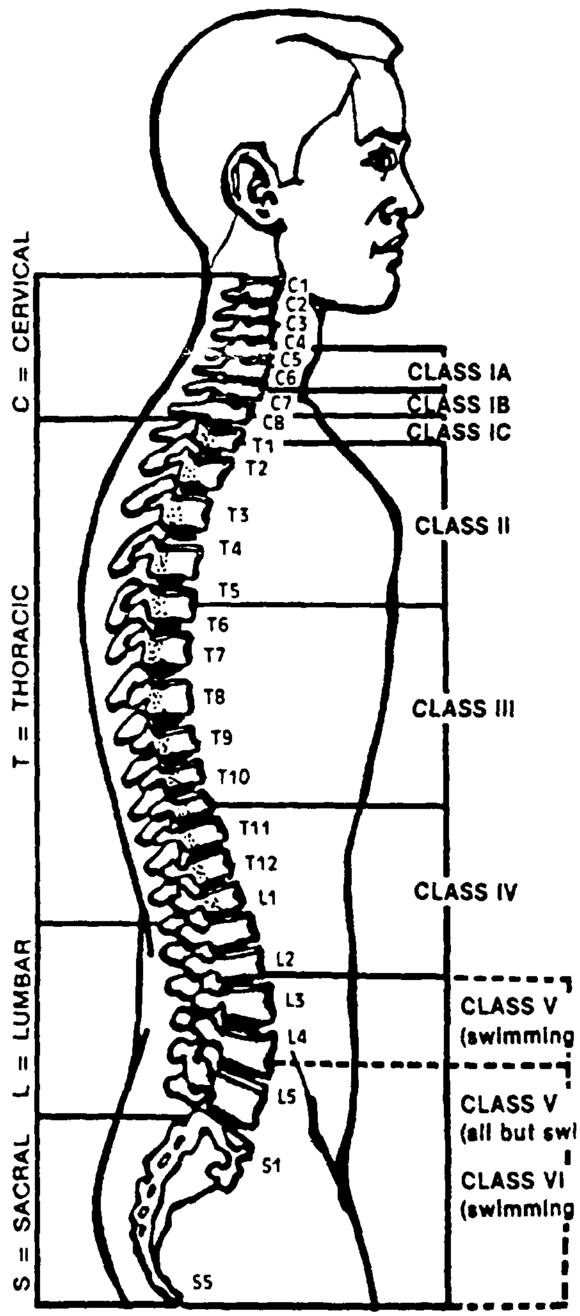

Figure 9.
A graphic explanation of the classification system used by the National Wheelchair Athletic Association. (*Permission granted.*)

versy stems from defining a significant difference in ability to propel a racing wheelchair or in performance. Coutts (22) has proposed the development of new methods of classifying athletes based on performance and evaluations by trained observers. His proposal would reduce the number of classes from seven to four (two quadriplegic and two paraplegic).

The ISMGF Athletics Committee is considering a proposal for a classification system that could be used to allow wheelchair athletes of all types (amputee, cerebral palsy, and spinal cord injured) to compete together. In addition, the proposal would reduce the number of paraplegic classes from four to two. Under the new system, athletes would first be classified by a medical team who would place each athlete in a class based upon level of impairment. Secondly, a trained observer (e.g., coach, official) and an experienced athlete (with recent experience, but not entered in the same competition) would evaluate the athlete being classified during preliminary rounds. Then the athlete would be placed in the class where his/her stroke kinematics, range of motion, seating position, and any other factors were most similar.

A more scientific basis for classification could involve a standardized exercise test, in addition to the methods listed in the previous paragraph. Athletes could be given a standardized maximal exercise test and/or anaerobic power test to help determine the "best" classification. The greatest foreseeable problem with this method (besides time and cost) is the high dependence upon training which may outweigh any disability-related factors.

BIOMECHANICS OF RACING WHEELCHAIR PROPULSION

The study of the biomechanics of racing wheelchair propulsion is a fairly new interest. Most of the published research appeared after 1980, and was based on the use of high-speed film.[3] Some of the tests were conducted in the laboratory while others were conducted on running tracks.

A typical four-link (wrist/hand, lower arm, upper arm, trunk) kinematic model used in biomechanical analyses of wheelchair racing is shown in **Figure 10.** The most commonly cited kinematic analyses are those of Higgs (43), Ridgway, *et al.* (61), and Sanderson and Sommer (62). **Figures 11a** and **11b** show typical joint trajectories during steady state in Cartesian coordinates. Each of these studies and a recent study by Cooper (13) have investigated the cycle time (the total time for each stroke). The percent of the cycle time spent in propulsion and recovery for each of these studies is presented in **Figure 12.** The mean percent cycle time at about 6 m·s^{-1} for all of these studies spent in propulsion was 36.25 percent, and in recovery was 63.75 percent.

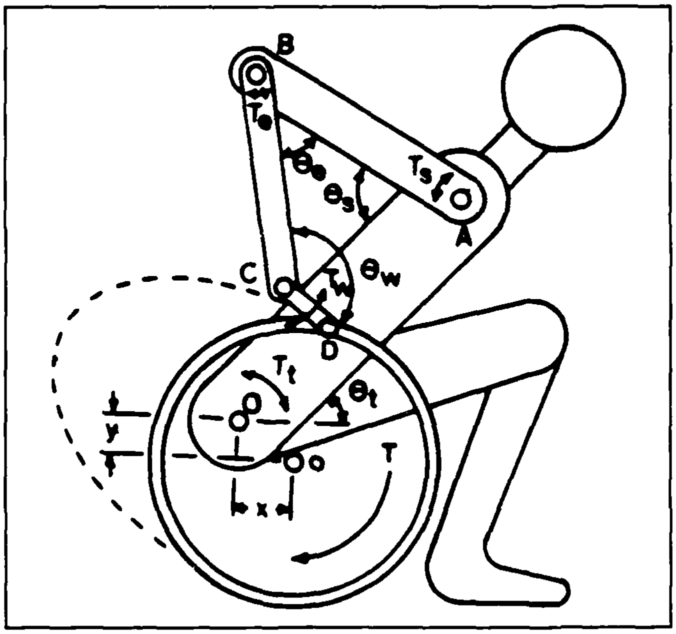

Figure 10.
A four-link kinematic model for biomechanical analysis of racing wheelchair propulsion.

The results of Ridgway, *et al.* (61) show that different levels of injury have distinct differences in their stroke kinematics with more potentially-able athletes (athletes of higher class) exhibiting less head movement (Class II/III showed 13.9 degrees while Class IV/V showed 9.2 degrees) and greater trunk movement (Class II/III showed 3.45 degrees while Class IV/V showed 7.8 degrees) during propulsion. This is probably a result of athletes attempting to generate greater propulsive force at the push-rings by imparting some momentum from their trunk to the push-rings. The more severely mobility-impaired paraplegics show more head movement because of the reduced functional control of the trunk, thus transferring momentum to the push-rings via head movement. In addition, the thighs of the Class IV/V athletes were positioned further from a vertical reference line than the other classes (IA/IB 32.18 degrees, II/III 37.60 degrees, IV/V 50.30 degrees).

Van der Woude, *et al.* (72,73) have suggested that the stroke kinematics are related to push-ring diameter, speed, work load, and fitness of the athlete. There is some indication that various push-ring diameters result in different demands on the athlete's body, and that optimum shoulder position is a function of push-ring diameter and the level of fitness of the athlete. They also found that the energy expenditure increases with push-ring size for the range from

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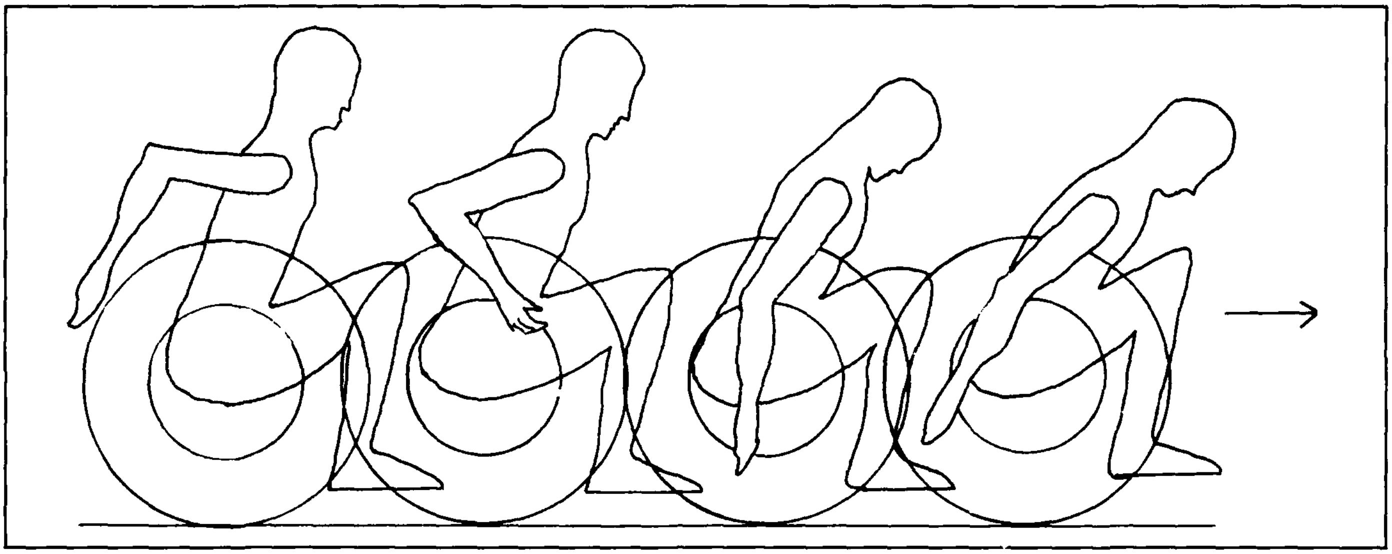

Figure 11a.
A representation of the arm movements during wheelchair racing.

0.3 to 0.56 meters of racing wheelchair push-ring diameters. A maximum efficiency of 6.7 percent was obtained for the large push-ring as compared to 7.9 percent for the smallest push-ring at speeds of 2.5 $m \cdot s^{-1}$. These results may be biased by the push-ring diameter the subjects are accustomed to, and by the position of each individual's shoulders with respect to the push-rings.

Walsh, *et al.* (74,75) have suggested that the ability to achieve higher pushing frequency during the sprint start is related to the rate of acceleration, and in the acceleration phase the path of the hand during the follow-through should be along the push-ring. They found that during acceleration their subjects used cadences of between 100 to 150 strokes per minute; at constant speeds, cadences of between 50 and 80 strokes per minute were observed. The difference in power transmission for various push-rim and wheel diameters is often not measured or reported. This may be done by calibrating the ergometer, dynamometer, or treadmill for power output and then accounting for the gear ratio of the wheelchair. This could be as simple as reporting the workload, wheel diameter, and push-rim diameter.

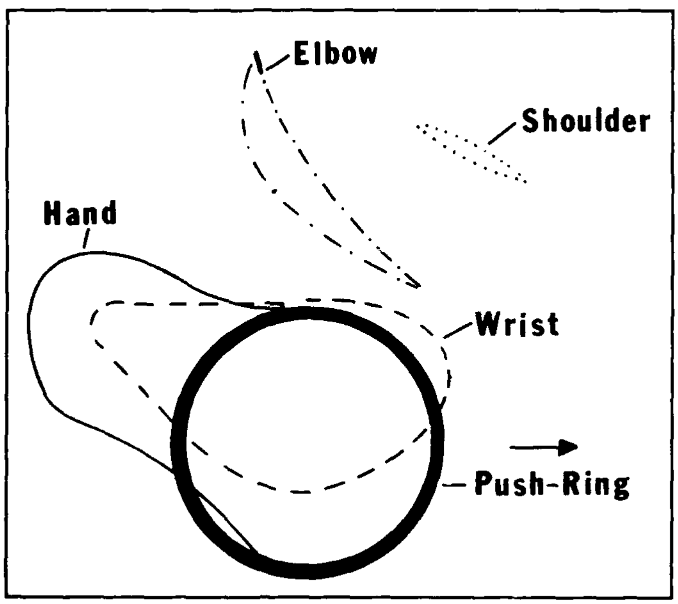

Figure 11b.
Example joint trajectories during racing wheelchair propulsion.

There are several factors which influence the determination of an optimum of such complex human/machine systems: 1) the effects of training or accommodation to the experimental station; 2) the adaptation of the subjects to changes made by the investigators; 3) the complex coupling of many factors (when one thing is changed it affects several other things), preventing the complete decoupling (separation so that we know the change was a direct result of the variable we are studying) of any one variable from another; and, 4) preventing the direct correlation of the change in one variable to the change in another. (How do we know the change was due to the factors we are studying and not due to something else, or is there a temporary effect that may improve, or a detriment performance that does not reflect the long term effect?) For example: if one proposes to determine the effect changes in push-ring size have on speed, the problem is complicated by the change in the shoulder position with respect to the various push-ring sizes. The anthropometric measurements of each indi-

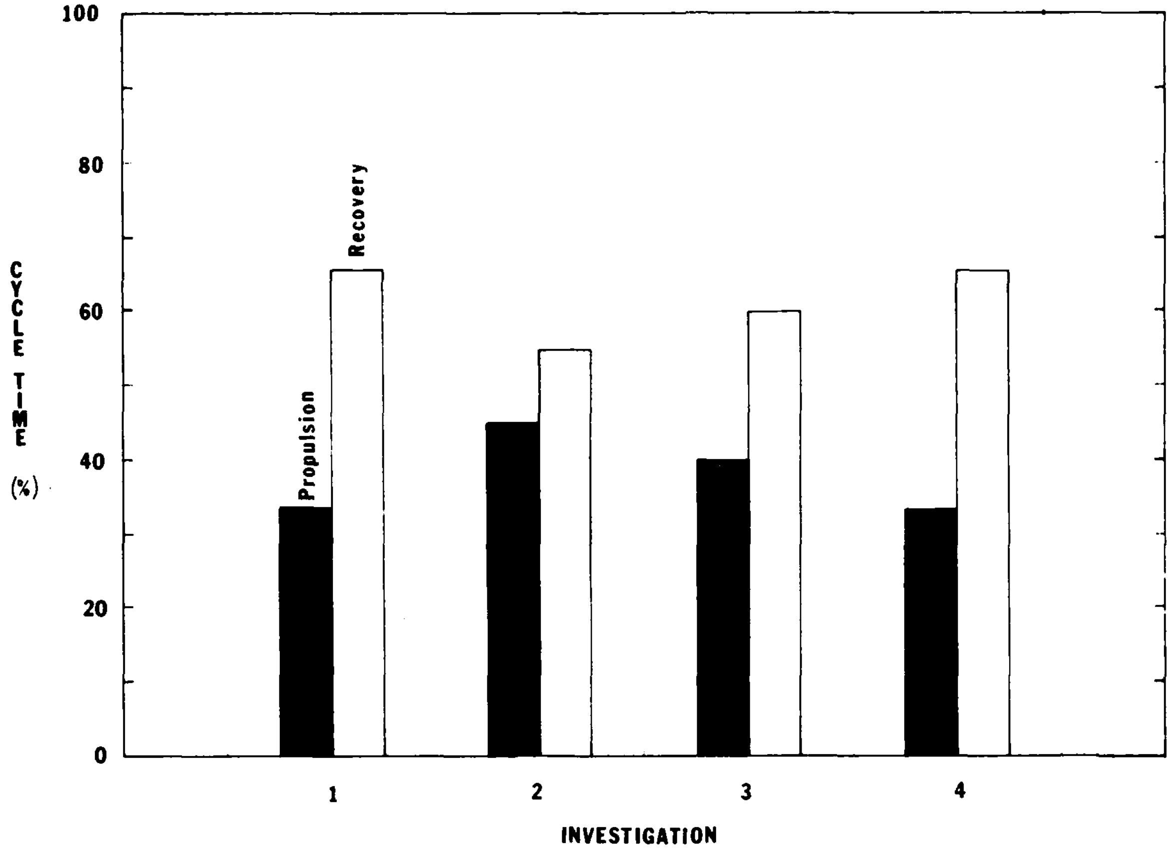

Figure 12.
Bar graph showing the percentage of the total cycle time spent in propulsion and recovery. (1 = Ridgway *et al.*; 2 = Higgs; 3 = Sanderson and Sommer; 4 = Cooper)

vidual influence the results as well as the effects of training (in general, an athlete will be most efficient with a pushring nearest the size used on his/her racing chair in the short term, but this may not be the athlete's optimum pushrim size).

Tupling, *et al.* (71) studied the efficiency of grab and strike starts and determined that grab starts are most efficient (grab starts produced an average impulse of 152.6 N•s, whereas strike starts produced an average impulse of 119.5 N•s). In addition, they determined that starting ability (impulse scores) were correlated to maximum strength ($r=0.89$, $n=8$).

Analyses of muscle fiber types for the prime movers have been conducted to a limited extent and suggest there may be a correlation between the ratio of the muscle fiber types and the potential to be competitive in a particular sport (68,69). The prime movers for wheelchair racing were identified to be the anterior deltoid, the pectoral, and the triceps muscles. Tesch and Karlsson (69) found that the wheelchair athletes they tested had 47 ± 12 percent fast twitch muscle fiber, and a ratio of fast twitch fiber area to slow twitch fiber area of 1.61 ± 0.44 with 58 ± 10 percent of the total area represented by the fast twitch muscle fiber.

The analysis of the athlete/wheelchair interaction has been limited because many factors could not be measured. High speed filming is limited because it does not give an accurate reflection of the actual propulsion cycle (only the point of contact can be estimated and not the point where a propulsive force is initiated), nor is there any information concerning the magnitude and direction of the forces.

Treadmill testing is limited because the recovery phase of the propulsion cycle is modified. Some investigators have attempted to circumvent this problem by filming the individuals on the track; however, this is also limited because when using a fixed camera position only a short window (about five meters) can be observed.

Evaluating the biomechanics of wheelchair propulsion has been limited by the available instrumentation and the apparent lack of coordination between investigators of different disciplines (13,26). Most investigators of manual wheelchair propulsion have no means of measuring the three-dimensional forces/torques impelled on the push-rim

by the individual. The California State University at Sacramento presently has a set of racing wheelchair wheels capable of measuring the three-dimensional forces and torques acting upon the push-rims. The Hines Department of Veterans Affairs Rehabilitation Research and Development Center is working with the University of Illinois at Urbana-Champaign to develop a set of wheels for a typical everyday-use wheelchair capable of measuring the three-dimensional torques and forces. Wright State University has a wheelchair ergometer capable of measuring push-rim torque.

Instrumentation needs to be developed to measure and record acceleration, speed, and work during actual racing or training. More sophisticated dynamometers need to be developed to simulate racing and training conditions. Instruments need to be developed to study the effects of wind, bearing, and rolling resistance.

TRAINING AND COACHING OF WHEELCHAIR RACERS

Training and coaching of wheelchair racers is probably one of the most neglected areas of wheelchair racing science and medicine. Little information has been disseminated on effective coaching and training techniques; in this light, the athletes' progress since the inception of wheelchair racing in 1961 is impressive.

With the growing interest in wheelchair sports, an interest rose in the effect different types of training had on wheelchair racing performance (21,23). Hedrick, *et al.* (40) have prepared a manual on wheelchair sports for the Paralyzed Veterans of America using techniques adapted from running and cycling, as well as experience accumulated at the University of Illinois, Urbana-Champaign. The manual briefly covers seating and positioning, biomechanics, sports medicine concerns, nutrition, and training programs.

Dreisinger and Londeree (27) have written a review on wheelchair exercise that contains useful information on training for fitness. Steadward and Walsh (65) have published a review on wheelchair training strategies past, present, and future.

A few papers have been published on strength training,[4] showing it to be an important supplement to wheelchair training, and providing some useful training routines (30). Gross investigated the effect fitness training has on the strength and endurance of quadriplegic individuals, and found that regular training can increase respiratory fitness.(33) In general, most of the rules for training ambulatory athletes apply to wheelchair athletes as well. Clearly, upper-body strength is important to all physical sports, and aerobic training is required for success in endurance events.

A survey of the training practices of elite wheelchair road racers conducted by Hedrick, *et al.* (39) showed that most athletes rely on each other for information concerning training and that they follow no structured training program. They found that elite male racers train an average of 63 minutes per day for 6.8 workouts per week (74 miles per week) over the entire year, with the maximum amount of time spent training during April through June, and the minimum during October through December. Women train an average of 70 minutes per day for 5.7 workouts per week (54 miles per week) over the entire year, with the maximum amount of time spent training during October through December, and the minimum during April through June. Most athletes generally practice good health with regard to the use of tobacco, alcohol consumption, diet, and weight control (9). Few wheelchair athletes are coached formally, in contrast to able-bodied athletes, which suggests an area for future advancements in the sport (31).

As in any type of sport involving direct competition, strategy plays an important role in wheelchair racing. There has been very little published on this subject (64) and this lack of available information has had a noticeable effect on wheelchair race results. In international competitions, teams training regularly under experienced coaches often try to force the development of strategical races; then, by dictating the strategy, gain an advantage over other competitors.

Some research has been involved with the effect of nutrition on performance; preliminary results suggest this should be of greater concern to the wheelchair athlete.[5] These studies imply that wheelchair racers understand little about nutrition for competition (39,40). Nutritionists suggest that good dietary habits developed for able-bodied athletes also would be effective for most wheelchair racers.

The first works concerning the positive psychosocial aspects of wheelchair sports[6] were produced during the 1970s (4,54). Psychological profiles (the Profile of Mood State, and State-Trait Anxiety Inventory) of elite-level wheelchair racers and elite-level able-bodied athletes are similar (41,44,57).

Sports have been shown to have a positive influence on the psychological rehabilitation of the mobility-impaired by improving self-image and self-esteem, and by being an outlet for anger and frustration (34,35,36,45). Wheelchair sports and racing help mobility-impaired persons focus on the positive aspects of their lives rather than dwelling on the negative aspects; thus helping them adjust to the changes

in their lives (8). Wheelchair athletes tend to be better adjusted to their disability[7] than do their nonathletic counterparts (58,63). Sports are also a vehicle for improving the able-bodied population's perception of the wheelchair user. The abilities and natural desires of wheelchair users to succeed and be recognized for their accomplishments (on an equal basis with their able-bodied peers), along with the ability of some of them to meet or exceed the performances of their able-bodied peers[8] has earned them greater respect and understanding for their accomplishments, rather than for their courage and fortitude (37,38).

With the increased drive to succeed in sports, primarily in wheelchair road racing where substantial amounts of prize money are available, wheelchair athletes have become more interested in sports psychology and a small amount of information directed at them has been published. These athletes have many of the same concerns and goals as their able-bodied peers and use the same tools from sports psychology to improve their performance (i.e., visualization, focusing, concentration, techniques to relax, etc.) (1,2,3,32,51).

Organized wheelchair sports teams are still generally associated with rehabilitation centers (50,66,67). Although this is an effective means of introducing newly-injured individuals to wheelchair sports, it may not be the best situation for developing "elite" wheelchair athletes. At rehabilitation centers, resources to hire a trained, full-time coach specializing in wheelchair sports are often not available. Most experts in exercise physiology, sports medicine, sports science, and engineering are not associated with rehabilitation centers. This removes the wheelchair athlete from potentially valuable resources and distances the expert from the sport. A more scientific approach to training should be developed so that wheelchair athletes can achieve higher performances, and have long healthy athletic careers with a minimum risk of injury. The health benefits and risks of wheelchair racing and training need to be studied in greater detail and with a broader subject population. An injury register should be developed to make data on safe training and racing habits available to athletes, coaches, trainers, and physicians.

SUMMARY AND CONCLUSIONS

Future investigations of wheelchair racing science must take an interdisciplinary approach. There is much to be understood about the interaction between an athlete and his/her racing wheelchair and the interaction of the athlete/wheelchair system with the environment. These complex problems require an understanding of engineering, physiology, and biomechanics.

A substantial body of knowledge exists concerning racing wheelchair sports science and medicine. Investigations cover a wide range of topics, as is expected of any multidisciplinary subject. Much of the work is difficult to compare due to the lack of consistent procedures from one investigation to another. The data presented in the literature are often incomplete.

It is critical, when reporting results of experiments with wheelchair racing athletes, to report the size of the propulsion wheels, the size and shape of the push-rings, the type of contact surface used on the push-rings, and the type of contact surface on the gloves used. It is also important to include the dynamic location of the shoulder in relation to the rear axle in studies of the biomechanics of wheelchair propulsion and, most importantly, when varying any of the wheelchair parameters (i.e., the push-ring size, axle position, etc.). Studies need to be more consistent in their definition of a racing wheelchair and an elite athlete.

Wheelchair sport science is still in its adolescence, but a basis for future investigation has been established. Most of the investigations thus far have been concerned with studying male paraplegics (they constitute an overwhelming majority of wheelchair racers): future studies need to include female and quadriplegic wheelchair athletes.

Much remains to be investigated in all of the areas discussed in this review. Mathematical modeling and analysis, computer simulation, and the engineering of more sophisticated test equipment are crucial to the continued development of this field.

ACKNOWLEDGMENTS

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END NOTES

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ABSTRACTS OF RECENT LITERATURE

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Abstracts are drawn primarily from the orthotics, prosthetics, and sensory aids literature. Selections of articles were made from these journals:
American Journal of Otology
American Journal of Sports Medicine
Archives of Physical Medicine and Rehabilitation
Assistive Technology
British Journal of Audiology
Ergonomics
Foot and Ankle
IEEE Transactions in Biomedical Engineering
Human Factors
International Journal of Rehabilitation Research
Journal of the Acoustic Society of America
Journal of Bone and Joint Therapy
Journal of Medical Engineering and Technology
Journal of Prosthetics and Orthotics
Journal of Speech and Hearing Research
Journal of Visual Impairment and Blindness
Language Speech Hearing Services Schools
Orthopedics
Physical Therapy
Physiotherapy
Prosthetics & Orthotics International
Scandinavian Journal of Rehabilitation Medicine
Seminars in Orthopaedics

PROSTHETICS, ORTHOTICS, AND RELATED TOPICS

ABLEDATA and Hyper-ABLEDATA. Hall M, Vanderheiden G, Thompson C, reprinted from *Assist Technol* 1:107-111, 1989.

This article reviews the status and content of ABLEDATA, a database that indexes products useful to persons with disabilities. The various methods of accessing ABLEDATA are described, including the availability of Hyper-ABLEDATA, the new desktop version of the database for Macintosh computers. [JEE]

An Arm-Powered Racing Bicycle. Cooper RA, reprinted from *Assist Technol* 1:71-74, 1989.

A new arm-powered racing bicycle is described. The features and performance of this bicycle are delineated, and the relative advantages over tricycles are outlined. [JEE]

The Application of Neural Networks to Myoelectric Signal Analysis: A Preliminary Study. Kelly MF, Parker PA, Scott RN, reprinted from *IEEE Trans Biomed Eng* 37:221-230, 1990.

Two neural network implementations are applied to myoelectric signal (MES) analysis tasks. The motivation behind this research is to explore more reliable methods of deriving control for multidegree of freedom arm prostheses. A discrete Hopfield network is used to calculate the time series parameters for a moving average MES model. It is demonstrated that the Hopfield network is capable of generating the same time series parameters as those produced by the conventional sequential least squares (SLS) algorithm. Furthermore, it can be extended to applications utilizing larger amounts of data, and possibly to higher order time series models, without significant degradation in computational efficiency. The second neural network implementation involves using a two-layer percep-

tron for classifying a single site MES based on two features, specifically the first time series parameter, and the signal power. Using these features, the perceptron is trained to distinguish between four separate arm functions. The two-dimensional decision boundaries used by the perceptron classifier are delineated. It is also demonstrated that the perceptron is able to rapidly compensate for variations when new data are incorporated into the training set. This adaptive quality suggests that perceptrons may provide a useful tool for future MES analysis. [JEE]

Biomechanics and Shape of the Above-Knee Socket Considered in Light of the Ischial Containment Concept. Pritham CH, reprinted from *Prosthet Orthot Int* 14:9-21, 1990.

In recent years considerable interest has been generated in the United States and abroad about new style above-knee prosthetic sockets, variously referred to as Narrow M-L, NASNA, CAT-CAM and SCAT-CAM. More than a little confusion has attended the process. Moreover, the impression has been created that they are not governed by the basic biomechanical rules identified by Radcliffe as affecting the quadrilateral socket. Attention has come to be focused on the role of ischial containment and the term Ischial Containment (IC) socket is enjoying widespread use. This paper reviews many of the critical features of such sockets with the goal of first demonstrating that many of these features are dictated by the requirements of ischial containment, and second that the principles set forth by Radcliffe are fully applicable. The paper concludes with a brief discussion of the alignment principles associated with Long's Line. [JEE]

Body Load in Heel-strike Running: The Effect of a Firm Heel Counter. Jorgensen U, reprinted from *Am J Sports Med* 18:177-181, 1990.

The effect of a firm heel counter in the shoe was studied in 11 athletes during submaximal heel-strike running on a treadmill under standardized conditions. The runners were tested in identical shoes with and without the distal 2 cm of the firm heel counter. Body load was expressed by absolute and relative $\dot{V}O_2$, surface EMG on the right leg, and *g*-force registration from an accelerometer below the right tibial tuberosity.

The heel counter caused a 2.4% significant decrease in $\dot{V}O_2$, a reduction in musculoskeletal transients, and a decrease in the activity of the triceps surae and quadriceps muscles at heel strike.

The changes found are expressions of kinematic adaptations in the body to increased or decreased load and provide functional evidence for the loading factor in the pathophysiology of overuse injuries. [JEE]

CAD-CAM Applications for Spinal Orthotics: A Preliminary Investigation. Raschke SU, Bannon MA, Saunders CG, McGuiness WJ, *J Prosthet Orthot* 2:115-118, 1990.

The British Columbia Institute of Technology and the University of British Columbia collaborated on a study to determine the feasibility of applying Computer Aided Design-Computer Aided Manufacture (CAD-CAM) techniques to the thoracolumbosacral orthosis for treatment of nonstructural spinal curve. The system used CANFIT, marketed by Shape Technologies, Inc., for below-knee prosthetic sockets. CANFIT modified an original torso shape to match a hand-modified one, both done from the same cast of a child with idiopathic scoliosis, with right thoracic curve of 24 degrees, and left lumbar curve of 23 degrees. The CANFIT system was not suited to create large torso shapes. After conventional casting, the investigators modified the cast and made a pelvic section to conform the adequacy of modification. They also made a prosthetic foam model of the modified cast. From this model, they used a shape copier to enter the shape into a data file in the CANFIT system. They also input the model of the hand-modified cast, and superimposed the desired end result over the shape being modified to check the progress of modification. Modification consisted of changing the original shape input and comparing modifications with the fully modified shape, until the two matched. The general modification of CANFIT allowed creation of the torso shape; this function was originated to accommodate unusual variations in socket shape, such as those caused by bony spurs. The specialized below-knee modifications of CANFIT were useless on the torso shape. After modifications were completed, a numerically controlled milling machine carved a model of prosthetic foam.

Using the general modification function required setting the parameters for the area to be modified, determining the depth and location of the deepest point in the modification, then adding and removing material in adjacent regions.

The time-consuming process sometimes produced lumpy results. For spinal orthoses, other modifications should be added, such as centering the trunk for patients with lateral shift, altering the height of the pelvic crest for those with leg length discrepancy, and rotating the trunk over the pelvis.

Some modifications were easy, such as compression of the abdominal area, while buildups over the anterior superior iliac spines were difficult. The waist crease was the most difficult modification. The next step is development of software and hardware to eliminate problems identified in the feasibility study to create a spinal CAD-CAM system comparable to the CANFIT system. [JEE]

Clinical Evaluation of the Rocker Bottom Crutch. Basford JR, Rhetta HL, Schleusner MP, reprinted from *Orthopedics* 13:457-460, 1990.

One hundred fifty hospitalized patients referred to a physical therapy department for crutch walking instruction were evaluated in a randomized, controlled crossover study with "rocker bottom" and conventional axillary crutches. A large training effect was observed with each crutch, but no significant differences (all $P > .05$) of gait speed, stride length, heart rate, stability, or feeling of security were noted between the groups. Thus, rocker bottom crutches, despite potential stability and energy conservation benefits, were found to be no more effective than conventional axillary crutches in this hospital setting. [JEE]

Design Metholodology for Aids for the Disabled. Orpwood RD, reprinted from *J Med Eng Technol* 14:2-10, 1990.

Although many aids for the disabled appear to be quite simple, their design is subject to many difficulties. These problems arise because of the intimate relationship between the equipment and the human body. Equipment design is therefore affected by many biological variables that are difficult to define at the beginning of the design exercise and many will not even be obvious until a device is tried out with a user. Standard design methodologies rely on all the variables being defined in a thorough specification at the start of the design process and therefore often lead to ineffective devices or very long development programmes. An alternative methodology is presented which overcomes the problems by separating the user interface aspects of the design from the supporting features. The user interface aspects are then allowed to evolve in conjunction with tests with potential users and finally integrated with the supporting features once a satisfactory solution has been found. Much more effective devices result with much shorter development effort. [JEE]

Development of a Realistic Method to Assess Wheelchair Propulsion by Disabled People. Mattison PG, Hunter J, Spense S. *Int J Rehabil Res* 12:137-145, 1989.

Ten individuals with cerebrovascular accident, who propelled a wheelchair with one arm and one leg, and six persons with vascular disease, paraparesis or amputation, who propelled a wheelchair with two arms, maneuvered their wheelchairs around a test circuit on a level floor. The circuit consisted of a large oval, and large and small figures of eight. Prior to rolling, they had their resting pulse rates recorded. They moved at preferred speed for 12 minutes or until feeling discomfort. Pulse rate was monitored continuously. Distance and time were recorded, and the physiological cost of propulsion was calculated from pulse rates and speed. Subjects also rated the extent of exertion. The performance of disabled subjects was compared with that of ten nondisabled volunteers.

Among the disabled subjects, speed varied from 0.23 to 0.8 meters per second, and pulse ranged from 8 to 50 beats per minute. Physiological cost was not related to age or diagnosis, but did reflect effort expended by each participant. Physiological cost was unrelated to recovery time, suggesting that recovery reflects cardiorespiratory fitness rather than exertion. Physiological cost correlated significantly with the subject's perceived exertion. Mode of wheelchair propulsion did not affect distance, speed, physiological cost or perceived exertion. Eleven subjects also used an arm crank wheelchair; ten traveled farther, and eight had reduced physiological cost. [JEE]

The Effect of an Abdominal Belt on Trunk Muscle Activity and Intra-abdominal Pressure during Squat Lifts. McGill SM, Norman RW, Sharratt MT, reprinted from *Ergonomics* 33:147-160, 1990.

The purpose of this study was to determine whether abdominal belts such as those prescribed to industrial workers reduced trunk muscle activity and/or increased intra-abdominal pressure (IAP). In this study, six subjects lifted loads (72·7 to 90·9 kg) both with and without wearing a weightlifter belt. In addition, further trial conditions required that subjects lifted both with the breath held or continuously expiring on lifting effort. Dynamic hand loads were recorded together with intra-abdominal pressure (IAP) and abdominal, intercostal and low back EMG. Every subject demonstrated an increase in IAP when wearing the belt during both breathing conditions: 99 mmHg with no belt; 120 mmHg wearing belt ($p < 0{\cdot}0001$). However, it was also found that significant increases in IAP occurred

($p < 0\text{·}017$) when the breath was held versus exhaling with or without the belt. One would expect that if the belt relieved either the direct compressive load on the spine or assisted IAP to produce an extensor moment then this would be reflected in diminished extensor muscle activity. Erector spinae activity tended to be lower with the breath held, suggesting a reduced load on the lumbar spine, although wearing a belt did not augment this reduction. In the case studies with subjects wearing an ergogenic corset designed for use by industrial manual materials handlers, perceptions of improved trunk stability were reported. However, the muscle activity and IAP results of this study during short duration lifting tasks make it difficult to justify the prescription of abdominal belts to workers. [JEE]

The Effect of a Viscoelastic Orthotic on the Incidence of Tibial Stress Fractures in an Animal Model. Milgrom C, Burr DB, Boyd RD, et al., *Foot Ankle* 10:276-279, 1990.

Ten mature rabbits weighing 3.3 to 4.5 kg, who had hindlimbs in moleskin-lined splints, were compared with 12 rabbits who had a splint lined with 3mm-thick viscoelastic material. The rabbit was in a tray with legs protruding through cutouts. The right hindlimb splint was attached to a cam follower. As the cam rotated, the limb was loaded by the inertia of the tray. The leg limb was splinted, but not loaded. Nonpainful loading was 1.5 times body weight during daily 40-minute treatments. Repetitive impact loading was over a 50-millisecond interval at 60 cycles per minute. Prior to loading, animals had X-rays and bone scans, which were repeated periodically. After final loading, the activity of the hindlimbs was imaged. Three days later, the animals were sacrificed, and the tibiae removed and X-rayed.

There were no statistically significant differences in rate of fracture among those with and without viscoelastic splints; 90 percent of the former and 92 percent of the latter had stress fractures. The animal model is analogous to the human situation because stress fractures in both occur in the diaphysis, there were cases of multiple stress fractures, not all subjects had fracture, bone scan activity progressed with training, and all grade 3 and 4 activity had X-ray evidence of fracture. The difference in model is that rabbits had passive loading. The inability of viscoelastic orthoses to prevent fracture may relate to the most important cause of fracture being bending force, whereas viscoelastic material attenuates compression loads. [JEE]

Energy Cost of Paraplegic Locomotion Using the ParaWalker–Electrical Stimulation "Hybrid" Orthosis. Nene AV, Patrick JH, *Arch Phys Med Rehabil* 71:116-120, 1990.

Five young men with complete paraplegia between T4 and T7 were proficient users of the hybrid ParaWalker. The orthosis consists of a rigid body brace, low-friction hip joints permitting flexion, and knee-ankle-foot orthoses with bail release knee-locks and limited ankle motion. The orthosis permits reciprocal gait. Subjects also used a two-channel portable electrical stimulator with electrodes on the gluteal muscles. Oxygen consumption was measured with the Oxylog. Subjects walked at steady pace at preferred speed while minute oxygen consumption and distance were recorded for 5 minutes. After resting, subjects had surface electrodes applied, then walked with the hybrid orthosis for 5 minutes.

Subjects traveled from 53 to 81 meters in 5 minutes without stimulation, and from 51 to 81 meters with stimulation. Three subjects increased walking speed by 8 to 11 percent with stimulation, while the other two reduced speed by 4 and 9 percent, respectively. Four subjects decreased energy consumption as much as 11 percent. Energy consumption ranged from 3 to 4 times resting consumption. Electrical stimulation prevented relative adduction of the stance leg which would have prevented the swing leg from passing freely. Stimulation of large gluteal muscles increases the active muscle mass, which may increase maximal aerobic capacity and anaerobic threshold to increase the capacity for sustained activity. [JEE]

Energy Cost of Walking with Hip Joint Impairment. Gussoni M, Margonato V, Ventura R, Veicsteinas A, reprinted from *Phys Ther* 70:295-301, 1990.

The energy cost of walking was measured in 12 patients (age 39-73 years) with hip joint impairments and 10 healthy controls during unassisted walking (2-6 $km{\cdot}h^{-1}$) on a level treadmill surface and on a 5% incline. The energy cost of locomotion in most patients increased up to 50% and 70% during level-surface and uphill walking, respectively. This difference between patients and controls was probably due to the increased external mechanical work. The energy cost of walking, although related to pain experienced during walking but not to hip joint range of motion or to joint status evaluated radiographically, provides an additional variable when defining the conditions of disability and functional impairment in individuals with this pathological condition. [JEE]

Falling Among the Sensorially Impaired Elderly. Tobis JS, Block M, Steinhaus-Donham C, et al., *Arch Phys Med Rehabil* 71:144-147, 1990.

Three populations were compared, namely: a) 60 independently functioning older adults who did not use walking aids; b) 58 independently ambulating blind older adults; and, c) 47 deaf subjects who ambulated without walking aids. Each population was divided into those older and those younger than 75 years of age. Nonimpaired and blind subjects were interviewed by premedical students, while the deaf subjects were interviewed by a social scientist using American sign language. All were interviewed in the presence of their peers. Interviews included questions pertaining to falls within the past year and the past 3 months, and health complaints and medication regimen. The reliability of the interview was demonstrated.

Deaf and nonimpaired subjects did not have a significantly different frequency of falling. Blind subjects were significantly different. Age younger or older than 75 had no significant effect on falling among the blind or deaf. Blind subjects reported falling outside the home for 54.8 percent of the group. Deaf subjects had only 31.3 percent who fell outside the home, while nonimpaired persons had 66.7 percent who fell outside. Blind fell because of tripping more often than feeling dizzy or no reason. Among the deaf, dizziness was the most common cause. Nonimpaired subjects fell primarily because of tripping. The data support the hypothesis that elderly fallers rely on visual information, rather than on kinesthetic and vestibular cues. Consequently, the visually impaired showed no increase in falling frequency with age, while the other two groups did increase falling with age. [JEE]

Functional Neuromuscular Stimulation for Standing after Spinal Cord Injury. Yarkony GM, Jaeger RJ, Roth E, et al., *Arch Phys Med Rehabil* 71:201-206, 1990.

Twenty-one patients, aged 18 to 50 years, with paraplegia, but no contractures or pressure sores, had quadriceps stimulation, followed by stimulation while standing in a frame with the hips hyperextended. The exercise stimulator was a custom-made variant of the standing stimulator. For exercise stimulation, parameters were pulses of 0 to 120 volts, 0 to 120 mA, 20 Hz frequency, and 0.4 ms pulsewidth. The standing stimulator has a postural change button which is depressed: 2 seconds elapse while the patient positions the hands for standing, an audible warning sounds, and stimulation begins. Sitting requires reversal of the action, whereby the same button is depressed, a delay ensues, an audible warning sounds, then stimulation amplitude decreases over a 2-second period. Carbon rubber electrodes are strapped to the thighs. Subjects stood on a force platform used to measure forces exerted on the ground.

Six subjects could stand without the assistance of a therapist outside the laboratory; they usually stood 5 to 15 minutes daily, 2 to 5 days a week. They were in the program for at least 6 months. The other subjects needed supervision often because they had entered the program recently and could not obtain adequate stability, or that they lived far from the laboratory. No subject had burn, increase in spasticity, or other complication. The system requires only simple equipment, without surgery or custom-made orthoses. The system does not provide consistent standing for more than 15 minutes, because of muscle fatigue. [JEE]

Gait Analysis in Clinical Practice. Gage JR, Ounpuu S, *Seminars in Orthopaedics* 4:72-87, 1989.

Gait analysis should not have encumbering equipment, should have a system which can deal with excessive motions in all planes, should not distract the subject unduly, and should have rapid analysis with data presented clearly: a single measurement tool is sufficient to understand neuromuscular deviations. Observational gait analysis cannot account for rapid events, nor activities occurring simultaneously at several joints: gait compensations are difficult to observe. Videotape, particularly slow motion with color and computer-controlled frame freeze, yields more consistent outcome, especially for knee flexion. However, assessment is qualitative. Temporal and stride measurements are easy to obtain means of quantifying gait outcome, but do not provide information about the cause of deviation. Foot switches and electrogoniometers are direct measurement devices which present problems of application to the patient. Switches are inappropriate for cerebral palsy patients who scuff their feet. Electrogoniometers are bulky and do not indicate position in relation to fixed spatial coordinates. Pressure can be measured by pressure-sensitive papers, pedobarograms, and force plates. Electromyography, whether surface or indwelling electrodes, presents data which is difficult to interpret because the signal can be confounded by electrode malposition, size, shape, interelectrode distance, subcutaneous fat, and muscle size, as well as the instrumentation.

Motion picture film analysis is time-consuming. Optical tracking, whether active marker with illuminated markers on the subject, or passive marker with reflectors on the

subject, is the most sophisticated method of data collection. Three markers are required on each body segment to define its three-dimensional motion. The system yields temporal and stride measurements, walking stick figures, and joint rotation graphs. Marker velocities and accelerations can be obtained.

A comprehensive gait analysis system which consists of a combination of various methods to measure all gait parameters, such as the system at Newington Children's Hospital, Newington, Connecticut, includes kinematic and kinetic data, electromyographs of 18 muscles, and temporal measurements, with results in 3 hours. Gait analysis is used routinely for preoperative decision-making and postoperative assessment. [JEE]

Guide Lines for Prescription of Walking Frames. Hall J, Clarke AK, Harrison R, reprinted from *Physiotherapy* 76:118-120, 1990.

This article describes the results, in terms of prescription guidelines, of a study which evaluated a range of commercially available walking frames. Twenty-eight frames, including pulpit, rollator and gutter frames were tested by regular users with the aim of identifying the good and bad design features. Results show that no one frame was superior but certain design features are important to consider when prescribing a frame, and the interaction of the user, frame and environment are fundamental. [JEE]

The Importance of the Toes in Walking. Hughes J, Clark P, Klenerman L, *J Bone Joint Surg* 72-B:245-251, 1990.

One hundred and sixty normal subjects, aged 5 to 78 years, had their foot positions recorded while standing and walking over a pedograph. The device consists of a thick glass plate with force transducers under each corner: the instrument also permits video recordings. The computerized system calculates the pressure distribution under 16 portions of the foot for each step. Peak pressure and force reached under each area, and the duration of ground contact were calculated.

Toes contacted the floor longer than the heel and base of the fifth metatarsal, but not as long as the metatarsal heads. Contact time was independent of age or weight. The hallux reached the highest peak pressure recorded under the foot. Central toes took more pressure than did lateral metatarsal heads. Pressures decreased from the first to the fifth toe. Peak pressure increased with body weight among children, but not adult subjects. Measurements taken during standing reveal lack of toe use. A third of the group had contact with all toes of both feet. A third did not use the fifth toe in standing, although 97 percent contacted the floor with the hallux, which takes 60 percent of the force under the toes. Peak pressure and contact time were unrelated to toe length; 80 percent had great toes longer than the second toes. Pressure and time were also unrelated to toe deformity, even though a fifth of the subjects had hallux valgus. Curly fifth toes were observed in 125 subjects. Half of the sample had plantar callosities, but there was no evidence that callosities were subject to pressures higher than the 95th percentile. Somewhat greater central metatarsal loading was present with increasing valgus angle, even in subjects without metatarsalgia. The prevalence of foot problems in the forefoot may be accounted for by the fact that the toes present a smaller area to sustain force during late stance. The toes contact the floor for about three-quarters of the walking cycle and exert pressures similar to those from the metatarsal heads. [JEE]

The Increase in Energy Cost of Walking with an Immobilized Knee or an Unstable Ankle. Mattsson E, Brostrom L-A, *Scand J Rehabil Med* 22:51-53, 1990.

Ten healthy adults participated in five walking tests. Heart rate was registered with a pulsometer. Walking speed was measured with a speedometer mounted on a cart pushed alongside the subject. Subjects walked at self-selected speeds. First, subjects wore Swedish clogs. Then the right foot had a clog with a convex sole designed to make the foot and leg unstable. The third test had subjects wearing ordinary clogs, but with the right knee splinted. The final two tests were performed with ordinary clogs, but at the speeds selected for the second and third tests. Energy was measured by a gas collection apparatus worn on the subject's back, connected to a mouthpiece.

With normal shoes, subjects walked an average of 78.7 meters per minute With an unstable ankle, speed decreased to 75.4, and with the immobilized knee, it dropped to 60.6 meters per minute. Optimum oxygen cost was 0.156 ml per kg per meter. With an unstable ankle, oxygen uptake was 0.169, a 10 percent increase. With normal shoes at the slower speed, uptake was decreased to 0.154. With knee splinting, oxygen uptake rose to 0.196 ml $\times$ kg $\times$ m, a 23 percent increase. With normal shoes at the slower speed, uptake decreased to 0.160 ml $\times$ kg $\times$ m. Consequently, knee immobilization increases energy cost significantly more than instability or normal walking. [JEE]

Isokinetic Strength Training in Below-Knee Amputees. Klingenstierna U, Renstrom P, Grimby G, Morelli B, *Scand J Rehabil Med* 22:39-43, 1990.

Eight men, aged 39 to 78 years, with below-knee amputation had isokinetic training of knee extensors and flexors. Six subjects had peripheral vascular disease. All wore a prosthesis daily and had full knee and hip excursion. Training was conducted for 8 to 12 weeks at angular velocities of 60, 180, and 240 degrees per second. Strength was measured before and after training. Biopsies from the vastus lateralis were measured with computerized tomography. Peak torque of the amputated leg increased significantly in all knee extension tests and in flexion at 180 degrees. The nonamputated leg also increased torque in most positions. Muscle fibers (especially type II) in the amputated leg, increased cross-sectional area in seven subjects, but the nonamputated side did not demonstrate increase.

Subjects stated they could walk twice as far after training and had less need of walking aids. [JEE]

Keyboard Emulation for Access to IBM-PC-Compatible Computers by People with Motor Impairments. Horstmann HM, Levine SP, Jaros LA, reprinted from *Assist Technol* 1:63-70, 1989.

The goal of this paper is to aid both clinicians and developers in understanding the issues associated with alternative input systems that permit full access to the IBM-PC family of computers. The first part of the paper discusses the concept of keyboard emulation in general and reviews a variety of keyboard-emulation systems that are currently available. The capabilities of a system call ALTKEY, developed in our laboratory, are described. The second part of the paper discusses the implementation of a keyboard-emulation system in more detail, using ALTKEY as a specific example. Technical design issues are discussed, and successful approaches to these design challenges are presented. [JEE]

Kinematics of High-Heeled Gait. Opila-Correia KA, *Arch Phys Med Rehabil* 71:304-309, 1990.

Fourteen women, aged 21 to 54 years, without lower limb fracture or previous surgery, were studied while wearing heels ranging from 5 to 7 cm, and with shoes having heels ranging from 0 to 2 cm. Kinematic data were collected with a five-camera Vicon motion analysis system with subjects walking at their preferred speeds. Subjects walked significantly slower with shorter strides and higher stance percentages with high-heeled shoes; cadence was unchanged. Significant differences were found in knee flexion and extension when comparing shoe heels. Flexion was greater at heelstrike and during stance phase, but was less at toe-off and during swing phase. Hip flexion during swing phase was also less with high-heeled shoes which also were associated with less pelvic tilt and less upper trunk motion.

With high heels, the hip was less abducted at toe-off; frontal plane motion of the hip also was less. The foot was more internally rotated throughout stance, while the hip was less externally rotated at heelstrike.

Intrasubject variability in each shoe type was computed. The lowest variabilities were in tibial and knee rotations in all planes where nine or more subjects had significant differences, although only six had differences in maximum knee flexion during stance. Hip and pelvic motions in the sagittal plane and trunk transverse rotations were highly variable in repeated trials of the same individual. Average variabilities for temporal parameters of cadence, stride length, and velocity were less than 3.4 percent. Quantified results correspond to visual observations with regard to the shortened stride, inverting and everting foot instability, increased knee flexion, and reduction in smoothness of gait; however, the expected exaggeration of pelvic and upper trunk motion was not found, perhaps because the shorter stride length creates the illusion of exaggerated trunk motion, but actually results in less pelvic and trunk motion. [JEE]

Measurement of Plantar Pressure Using Fuji Prescale Film: A Preliminary Study. Ralphs G, Lunsford TR, Greenfield J, *J Prosthet Orthot* 2:130-138, 1990.

Fuji Prescale film records pressure by color change. One application of pressure produces the same shade of red on the film as multiple applications. Pressure can be measured from 5 to 700 kg/cm^2. Eight women and two men who had complained of foot pain or had plantar callosities were studied. Their ages ranged from 23 to 51 years old and they weighed from 135 to 215 pounds. Their footcasts were modified by building up the area under the metatarsal heads. Custom cork and Spenco inserts were fabricated, as well as a second Plastazote insert. Fuji film was placed in the anterior portion of the shoe. Two strips of film were taped to the foot. One strip was a color-forming A-film, and the other was color-reactive C-film. The subject walked in the shoe with the film, but without any insert. The film patch was removed and the shade of red compared with a color-pressure chart. A second patch of film was applied

and the subject walked 30 steps with the cork/Spenco insert. The patch was removed and read; the procedure was repeated with the film and a one-quarter-inch Plastazote insert.

Mean pressures were 34.2 kg/cm^2 when walking without an orthosis, 15.9 kg/cm^2 with cork/Spenco orthosis, and 16.5 kg/cm^2 with Plastazote. [JEE]

Modification of a Proportional Joystick to Incorporate Switch Outputs for Accessories. Chizinsky KA, Horstmann HM, Levine SP, Koester DJ, reprinted from *Assist Technol* 1:101-105, 1989.

Integrating the control of accessory devices into a wheelchair joystick can both optimize function and improve aesthetics. Although the number of commercial systems having this feature is increasing, there is still a number of joystick controllers in the field that do not provide integrated accessory device control. This paper describes the design of an add-on system that allows the user to control both the wheelchair and accessory functions using a proportional joystick controller. The mode of operation is chosen with an external switch that can be mounted at any desired location. Accessory control can include both wheelchair functions, such as power recline, and external functions, such as an environmental control unit. The design is especially well-suited as a retrofit to existing wheelchair control systems. A case report of an individual with quadriplegia is presented demonstrating a successful implementation of this system. [JEE]

Myoelectric Hand Orthosis. Benjuya N, Kenney SB, *J Prosthet Orthot* 2:149-154, 1990.

A wrist-hand orthosis was designed for individuals with spinal cord injury at the C5-6 level, and has been fitted to two patients with complete C6 injury, and two with brachial plexus lesion. The handpiece of the orthosis is made of thermoplastic and supports the metacarpal arch and stabilizes the thumb in opposition to the index and middle fingers. A fingerpiece from the gearbox on the handpiece guides the stabilized fingers. The forearm piece houses the surface electrodes, a miniature motor, 6-volt nickel-cadmium battery, and a control circuit. The patient can don the orthosis using a hook on the forearm piece. The user then turns on the orthosis and activates it at will. The motor is controlled by circuitry which provides motor control proportional to electromyographic amplitude based on pulse-width modulation. A flexible shaft connects the motor shaft and the worm gear on the handpiece. Fingers close with a pinch force of 6 to 7 pounds and open to 10 cm within 6 seconds. Pinch force is displayed on a 5-LED array mounted on the handpiece so the user can distinguish mid-ranges of pinch force. The entire orthosis weighs 500 grams. The battery provides at least 8 hours of device use under normal operation.

A one-site electrode option functions such that when the muscle contracts, the hand closes, with speed proportional to intensity of myoelectric signal; relaxation opens the hand and brief contraction halts the hand at the desired opening position. The two-site option is arranged so that one electrode sends either a closing or an opening signal, and the other electrode sends the opposite signal; no muscular contraction maintains the position reached by the fingers so that the user can sustain pinch force without muscular contraction. Both options provide the same pinch power which is proportional to muscular contraction. Clinical trials with patients with spinal cord injury indicated they required as little as 10 minutes of training in order to activate the orthosis with minimal conscious effort. Friction gloves worn over the handpiece enable the patient to propel the wheelchair. One patient with brachial plexus injury had the electrode placed over the pectoralis major; another used signals from biceps and triceps on the uninjured arm. [JEE]

Oxygen Consumption and Cardiac Response of Short-Leg and Long-Leg Prosthetic Ambulation in a Patient with Bilateral Above-Knee Amputation: Comparisons with Able-Bodied Men. Crouse SF, Lessard CS, Rhodes J, Lowe RC, *Arch Phys Med Rehabil* 71:313-317, 1990.

Oxygen consumption of three nondisabled men of similar age was compared to a 37-year-old man with traumatic above-knee amputation. The patient walked with custom-fit short nonarticulated prostheses, 40 cm high, with long prostheses having Henschke-Mauch Swing-N-Stance hydraulic knee units and graphite-composite flex foot units. The patient could walk and run with the short prostheses without assistance, but used a cane with the long prostheses. All participants walked on a treadmill with a continuously monitored electrocardiogram and respiratory gas collection system. Subjects walked until exhaustion, starting with 37.8 meters per minute with progressive increase in treadmill velocity.

Energy cost was virtually the same with short and long prostheses; heart rate and minute ventilation also did not differ significantly. Treadmill exercise time, however, was 27 percent longer with the short prosthesis. The subject could walk at faster speeds with the short prostheses.

Compared with able-bodied subjects, the amputee used 47 percent more oxygen with short prostheses, and 79 percent more with long ones. Minute ventilation was considerably higher. Heart rate was also much higher: long prostheses required 141 beats per minute, short ones 124 beats per minute, and controls needed only 93 beats per minute. The amputee's measured oxygen consumption was 56 percent below his age-predicted value. [JEE]

Prosthetic Rehabilitation of Elderly Bilateral Amputees. Wolf E, Lilling M, Ferber I, Marcus J. *Int J Rehabil Res* 12:271-278, 1989.

This retrospective study of 18 patients who had bilateral amputation because of vascular disease indicates that the mean age of the 13 men and 5 women was 66 years. Six of the 12 patients with bilateral below-knee prostheses are functional with prostheses; the other bilateral below-knee amputees use wheelchairs. Those using prostheses did not have other diseases except an old myocardial infarction, diabetes, or cataracts. Those who could not walk with prostheses had congestive heart failure, stroke, or joint disease. The average time between amputations was 19.4 months. All those using two prostheses were successful prosthesis users after the first amputation. No one with one or two above-knee amputations wore a pair of prostheses. The most frequent reason for rejecting prostheses was poor effort tolerance rather than age. Associated diseases, level of amputation, and previous prosthetic use are key determinants of bilateral prosthetic use. Prosthetic rehabilitation is costly and requires much time and effort from the patient and staff. [JEE]

The Role of the Contralateral Limb in Below-Knee Amputee Gait. Hurley GRB, McKenney R, Robinson M, et al., reprinted from *Prosthet Orthot Int* 14:33-42, 1990.

Very little quantitative biomechanical research has been carried out evaluating issues relevant to prosthetic management. The literature available suggests that amputees may demonstrate an asymmetrical gait pattern. Furthermore, studies suggest that the forces occurring during amputee gait may be unequally distributed between the contralateral and prosthetic lower limbs. This study investigates the role of the contralateral limb in amputee gait by determining lower-limb joint reaction forces and symmetry of motion in an amputee and non-amputee population. Seven adult below-knee amputees and four non-amputees participated in the study. Testing involved collection of kinematic coordinate data employing a WATSMART video system and ground reaction force data using a Kistler force plate. The degree of lower limb symmetry was determined using bilateral angle-angle diagrams and a chain encoding technique. Ankle, knee and hip joint reaction forces were estimated in order to evaluate the forces acting across the joints of the amputee's contralateral limb. The amputees demonstrated a lesser degree of lower limb symmetry than the non-amputees. This asymmetrical movement was attributed to the inherent variability of the actions of the prosthetic lower limb. The forces acting across the joints of the contralateral limb were not significantly higher than that of the non-amputee. This suggests that, providing the adult amputee has a good prosthetic fit, there will not be increased forces across the joints of the contralateral limb and consequently no predisposition for the long-term wearer to develop premature degenerative arthritis. [JEE]

Shock Absorbing Material on the Shoes of Long Leg Braces for Paraplegic Walking. Bierling-Sorensen F, Ryde H, Bojsen-Moller F, Lyquist E, reprinted from *Prosthet Orthot Int* 14:27-32, 1990.

A study was designed to evaluate if shock absorbing material ethyl vinyl acetate (EVA) on the shoes of long leg braces could decrease the accelerations and consequent shock forces transmitted through the leg and brace during paraplegic walking. Six male paraplegics (26-55 years old) took part, four using a "swing-to" and two a "swing-through" technique when walking. Recordings comprised accelerometry of leg and brace, force platform measurement, and still photography of the trajectories of the leg segments. Each experimental condition was tested three times with a coefficient of variation (CV) for the measurements ranging from 5-22%. Compared to hard heels, shoes equipped with 20mm EVA soles decreased the acceleration amplitude in the first 10 msec as well as at maximum for shoe-to-ground contact. With the accelerometer at the malleolus reduction of the amplitude averaged 22% and 2% respectively, and 35% and 21% respectively with the accelerometer on the caliper (p: 0.03-0.1). In a second trial the two "swing-through" walkers had new shoes made with a 10mm thick EVA heel built in. After 3 months of walking with these shoes tests were carried out with the accelerometer attached to the malleolus both when the new and the former shoes were put on the calipers. CV for these measurements were 15-24%. It was found that the new shoes decreased the amplitudes by up to 62% and 26% on average (all $p<0.01$). The experimental subjects indicated that the EVA soles/heels gave a more comfort-

able and silent walk, e.g., the "bump" transmitted up through the body to the head diminished. In future, shock absorbing material should be built into the heels of shoes provided to long leg braces for paraplegic walking. [JEE]

Test Instrument for Predicting the Effect of Rigid Braces in Cases with Low Back Pain. Willner SW, reprinted from *Prosthet Orthot Int* 14:22-26, 1990.

The difficulty of predicting the acceptance and the result of wearing rigid braces has been identified before and is reported in the literature. Therefore a test instrument has been developed and tested. The intention is that the test instrument can imitate a rigid brace. Furthermore, different attributes of the rigid brace can be altered. Thus the range of the lordosis, the level of maximal dorsal support and the amount of abdominal support can be altered. By changing these parameters the maximal pain relief is sought. A good correlation between the result in the test instrument and the rigid brace manufactured according to the information from the former was seen (93%). No false negative results were seen. Thus, if no acceptance or pain relief was seen in the test instrument no pain relief could be expected in a rigid brace.

Another purpose of this test instrument is to simplify the manufacture of the brace and to transfer easily the information gained from the test instrument to the brace with the aid of a so-called measuring device. [JEE]

The Timing of Amputation for Lower Limb Trauma. Pozo JL, Powell B, Andrews BG, et al., *J Bone Joint Surg* 72-B:288-292, 1990.

The authors reviewed retrospective medical records of 35 patients having amputation because of severe trauma since 1975. All had surgery after failure of attempted limb salvage. The average age was 29 years, ranging from 6 to 80 years. Thirty-two were involved in traffic accidents, including 16 motorcyclists. Interval between injury and amputation ranged from 3 days to 16 years. All had either extensive soft-tissue loss with periosteal stripping and bone exposure or open fracture with arterial injury.

Those amputated within a month of injury had compound injuries with severe skin, bone, muscle, and vascular damage and wound contamination. This group of 7 patients had 6 above-knee and 1 below-knee amputation. The 13 who had amputations from 1 month to 1 year after an accident also had skin and muscle damage with bone loss or comminution and contamination, resulting in 7 above-knee and 6 below-knee amputations. Fifteen patients who had amputations, including 12 below-knee, after at least a year had severe skin, bone, and muscle damage and contamination. They averaged 12.3 operations.

Early management for all was debridement, stabilization, and vascular repair where possible. Salvage failed because of vascular injuries, often with ischemic failure, nerve damage, bone damage including nonunited fractures, muscle damage in three-quarters of the cases, insufficient skin cover, and sepsis. Those amputated early averaged a 15-week hospitalization. The others averaged longer hospitalization before amputation, and were discharged 6 weeks after amputation. Below-knee amputees were rehabilitated into a working environment in 6 months, as compared with 10 months for those with above-knee amputation. The study confirms the poor prognosis of severe compound injuries and indicates that even in the absence of neurovascular injury, the combination of severe skin loss, muscle damage, and bone loss with contamination is unlikely to lead to limb salvage. [JEE]

Trends in Finger Pinch Strength in Children, Adults, and the Elderly. Imrhan SN, *Human Factors* 31:689-701, 1989.

Pulp, chuck, and lateral pinches and grip force were measured in 62 children, 70 young adults, and 50 elderly adults with a Preston pinch meter and a Stoelting handgrip dynamometer. Two repetitions of each pinch and handgrip were taken with each hand. Stature, body weight, and hand breadth and length were measured. Hand laterality did not influence pinch or grip strength significantly. Except for young women, lateral grasp was strongest, followed by three-jaw chuck, thumb to index, thumb to middle, thumb to ring, and thumb to fifth finger. Women demonstrated more strength in chuck, averaging 7 kg, as compared with 6.5 kg in lateral grasp. Growth in pinch strength occurs at the same rate for different grasp patterns regardless of how frequently a pinch type is used or the muscle mass, length-tension properties, and leverage involved. Difference in pinch strength between age groups is constant, regardless of type of pinch. The weakest pinch, from thumb to fifth finger, was 27 percent the strength of the strongest lateral pinch, which was about a quarter the strength of handgrip. The average grip for young men was 49.7 kg, as compared with young women who averaged 31.4 kg. Elderly men gripped with 30 kg force, while older women produced 21.5 kg force. Ratios of various pinch and grip strengths were computed for each age group.

Absolute differences in strength between males and

females are slight in children, greatest in young adults, and somewhat less in the elderly. Hand size was a factor in pinch and grip only in children. [JEE]

Upper Limb Functions Regained in Quadriplegia: A Hybrid Computerized Neuromuscular Stimulation System. Nathan RH, Ohry A, *Arch Phys Med Rehabil* 71:415-421, 1990.

The system accepts voice commands through a recognition system connected to an Apple II microcomputer which controls a 24-channel stimulator which affects high-resolution, bipolar surface electrodes. The carbonized rubber electrodes concentrate the electric field activating the muscle beneath, without affecting adjacent muscles, until stimulation intensity increases to cause contraction by overflow of stimulation activity. Overflow may also be observed in distant muscles by reflex activity or excitation of nerves. The concentrated stimulation allows selectivity of activating the small hand muscles. No skin damage occurred after several hours of stimulation by current of 50 mA/cm^2 for square wave double pulses of 0.3 ms. Muscles innervated above the lesion can be activated voluntarily at full strength. Muscles innervated at the lesion do not respond minimally to stimulation, while those innervated below the lesion can be activated easily with electric stimulation. Most muscles could be stimulated directly, even deep ones which present superficial regions, except for extensor pollicis longus which could only be stimulated through extensor digitorum. Maximum strength of contraction depends on the accuracy of electrode placement. The ceiling on current intensity is reached when overflow occurs to other muscles.

Two subjects used the system, both with C4 quadriplegia. One needed an active wrist extension splint. Both had insufficient biceps and triceps force and thus used a suspension system. Visual feedback is presented on the computer monitor so that the user can make on-line adjustments to stimulation intensity. Audio feedback facilitates writing. The system recognizes 80 words, but is more efficient with fewer words. A vocabulary of 14 words enabled 80 to 90 percent recognition success rates. Three hand prehension and release programs have been programmed, each in 13 incremental steps to open the hand in the appropriate grasp, close the hand, open the hand to release the object, and relax the hand. The subjects could lift a 250 g cup to drink through a straw in it. One could feed herself using a fork attached to a long-angled handle. The system enabled both individuals to write.

Research is directed at simplifying the application of the system, now requiring 15 minutes. Splints and suspension are being redesigned and the stimulation and computing system hardware miniaturized. Alternatives to voice command are being investigated. [JEE]

Variability and Reliability of Joint Measurements. Bovens AMPM, van Baak MA, Vrencken JGPM, et al., reprinted from *Am J Sports Med* 18:58-63, 1990.

The purpose of this study was to determine the variability and reliability of joint measurements as carried out by three physician observers. The intratester variation and reliability of nine different joint measurements was determined in eight healthy subjects. The measurements were taken in eight sessions by each tester. In this population also the intertester variation and reliability was determined by the three observers. This was also done in a population of middle-aged athletes over a period of 2.5 years.

The results indicate that it is difficult to show either an improvement or worsening of a joint motion of less than 5 to 10 degrees for most joints measured by the same tester. The intertester variation is not consistent over a longer period of time, so differences between observers during long-term studies cannot be corrected on the basis of a single study at a single point in time. The reliability of all nine joint measurements is not very high, but is probably sufficient if the results are used to compare groups within a single population and for large studies with experienced observers. Because the reliability strongly depends on the interindividual variation, it is preferable to determine the reliability for each study population. [JEE]

SENSORY AIDS/REHABILITATION

A Comparison of Two Training Strategies for Speech Recognition with an Electrotactile Speech Processor. Alcantara JI, Cowan RCS, Blamey PJ, Clark GM, *J Speech Hear Res* 33:195-204, 1990.

Training over 70 hours with a multi-channel, electrotactile aid was given to 8 young adults by synthetic only (SO) or synthetic-analytic (SA) approach. Significant improvements in learning to recognize vowels, consonants, CNC words, and CID sentences, and in speech tracking occurred with both methods, but both did not yield reliable improvements on each measure. Authors conclude choice of procedure should depend upon trainee characteristics. [JDS]

Computer Simulation of the Patient for Training in Audiometry. Haughton PM. *Brit J Audiol* 24:45-50. 1990.

Criteria for a computer simulator for training students in audiometry are offered in detail. Author has constructed and tested a model that provides all features except central masking and that uses the computer keyboard in place of an audiometer: "No difficulties that would invalidate the model have been discovered." [JDS]

Hearing Loss, Aging, and Speech Perception in Reverberation and Noise. Helfer KS. Wilber LA. *J Speech Hear Res* 33:149-155. 1990.

Tests of speech perception by older and younger persons, with and without sensorineural hearing impairments, showed that age and amount of impairment are associated with hearing noisy, reverberant speech. However, age and HL were not significantly correlated with each other, though HL and consonant perception were. [JDS]

An Improvised Eye-Pointing Communication System for Temporary Use. King TW. *Lang Speech Hear Services Schools* 21:116-117. 1990.

Describes a device for presenting language materials to nonvocal patients that is made from a wire coat hanger, a plastic bag and some tape. Instructions for its use are included. [JDS]

Listener Experience and Perception of Voice Quality. Kreiman J. Gerratt BR. Precoda K. *J Speech Hear Res* 33:103-115. 1990.

Do naive and expert listeners differ in their judgments of voice quality? Recordings of 18 male speakers with voice disorders were rated by 5 naive and 5 expert listeners. Results show the two groups differed on bases of pertinence: naive listeners were similar in their strategies while clinicians differed widely among themselves. Authors conclude that caution should be observed in interpreting studies in which clinician's judgments are averaged. [JDS]

Magnitude of Diotic Summation in Speech-in-Noise Tasks: Performance Region and Appropriate Baseline. Davis A. Haggard M. Bell I. *Brit J Audiol* 24:11-16. 1990.

Testing of 27 persons with normal hearing revealed a 5 to 9 percent advantage for diotic over monotic hearing in noise, where *diotic* refers to presentation of identical waveforms to both ears, as opposed to *binaural* hearing, in which waveforms reaching each ear may differ. When hearing is identical in both ears, the use of the better monotic hearing level as the baseline for calculating the diotic advantage underestimates the latter by 3 percent or more. A method for estimating the 'true' advantage is proposed. [JDS]

Mental Health and Acquired Hearing Impairment: A Review. Jones EM. White AJ. *Brit J Audiol* 24:3-9. 1990.

The association between paranoia and impaired hearing has not been confirmed, nor has a correlation between acquired hearing loss and intelligence. However, persons with impaired hearing tend to have higher rates of social stress and depression, though the evidence supporting those claims suffers from lack of controls. 68 references. [JDS]

Mobility in Individuals with Moderate Visual Impairments. Long RG. Rieser JJ. Hill EW. *J Visual Impairm Blindn* 84:111-118. 1990.

The visual acuities, visual fields, contrast sensitivities, and videotaped performances of walking in unfamiliar indoor and outdoor routes, with normal and reduced lighting, were obtained from 22 adults with low vision. Ratings of the videotapes by orientation and mobility instructors were significantly correlated with contrast sensitivity (0.37) and visual field (0.38) but visual acuity was not. Based on a regression analysis, 30 percent of variance in mobility is accounted for by contrast sensitivity and visual field. [JDS]

Predictors of Tinnitus Discomfort, Adaptation and Subjective Loudness. Scott B. Lingberg P. Melin L. Lyttkens L. *Brit J Audiol* 24:51-62. 1990.

Data from a questionnaire administered to 3,372 persons drawn from all 69 hearing centers in Sweden—most of whom experienced both tinnitus and loss of hearing—showed interest in treatment was greater for those who complained about tinnitus than those whose major complaint was hearing loss, and greater for younger than older persons. Discomfort and adaptation to tinnitus were best predicted by degrees of control and of maskability by external sounds. [JDS]

A Procedure for Measuring Auditory and Audio-Visual Speech-Reception Thresholds for Sentences in Noise: Rationale, Evaluation, and Recommendations for Use. MacLeod A, Summerfield Q. *Brit J Audiol* 24:29-43, 1990.

Describes a lipreading test on videotape, consisting of 10 lists of 15 brief sentences calibrated to be equally intelligible, but whose results vary with presentation mode: only with hearing and with hearing supported by vision. Validation sample consisted of 24 persons. Complete list of test items and directions for use are in the article. [JDS]

Rapid Manual Abilities in Spasmodic Dysphonic and Normal Female Subjects. Cannito MP, Kondraske GV. *J Speech Hear Res* 33:123-133, 1990.

Finger-lift reaction time, index-finger tapping speed, and Purdue Pegboard Performance of 18 spasmodic dysphonic females were compared to controls. Controls performed significantly better on finger-lift and Purdue scores. Nondominant finger tapping speed and severity of speech impairment were significantly correlated. Potential for use in localizing the source of pathology is discussed. [JDS]

Relationships Between Selected Auditory and Phonatory Latency Measures in Normal Speakers. Stager SV. *J Speech Hear Res* 33:156-162, 1990.

Study of 34 normal speakers' laryngeal reaction times (LRT) supports an interaction between auditory and phonatory systems at the brainstem level. Authors raise questions about LRT measurement and propose an improved metric. [JDS]

Response Times to Speech Stimuli as Measures of Benefit from Amplification. Gatehouse S, Gordon J. *Brit J Audiol* 24:74-82, 1990.

Audiometric evaluations of 44 persons with impaired hearing, ranging in age from 51 to 80 years (mean=68 years), revealed that response time can be an effective measure of benefit. Authors describe an index based upon response times to single words and sentences that is particularly useful with 'problem' cases, where gains from amplification show ceiling effects and/or lie within the usual ranges of test-retest reliability. [JDS]

A Review of Technology-Related Publications. Leventhal JD, Uslan MM, Schreier EM. *J Visual Impairm Blindn* 84:127-130, 1990.

Describes and evaluates periodicals about assistive technology that are likely to be of interest to persons with visual impairments and in formats they can use: *BAUD, The Catalyst, Closing the Gap, Computer-Disability News, Newsbits, RDC Newsletter, SAINT, Tactic, Technology Update, VersaNews, Window on Technology.* [JDS]

Some Effects of Advanced Aging on the Visual-Language Processing Capacity of the Left and Right Hemispheres: Evidence from Unilateral Tachistoscopic Viewing. Rastatter MP, McGuire RA. *J Speech Hear Res* 33:134-140, 1990.

Based on a study of 24 persons 70 to 75 years of age, authors found stimulations of the right visual field gave significantly faster vocal reaction times than left field, and significant correlation between visual fields for abstract and concrete stimuli. These data are compared to earlier studies that support a ". . . callosal relay model of neurolinguistic organization." [JDS]

Some Effects of Signal Bandwidth and Spectral Density on the Acoustic-Reflex Threshold in the Elderly. Jakimetz JJ, Silman S, Miller MH, Silverman CA. *J Acoust Soc Am* 86:1783-1789, 1989.

Auditory-reflex thresholds (ART) of 20 young and 20 elderly persons with normal hearing decreased with spectral density to a plateau after 7 components for young and 5 components for elderly persons. Decreases were less for old than young persons. These results imply elderly persons' lesser ability to summate sound energy from complex signals and their poorer frequency resolution. [JDS]

Stapedectomy in Residency—The UAB Experience. Handley GH, Hicks JN. *Am J Otology* 11:128-130.

Review of 23 stapedectomies by residents revealed air-conduction closure relative to preoperative bone conduction of 20 dB or more in 9 cases, drop in bone conduction greater than 0-10 dB in 10, and complications in 5. Results improved with increase in number of surgeries performed. [JDS]

Tactual Environmental Interpretation: A Multisensory Approach. Nichols DR, *J Visual Impairm Blindn* 84:124-125, 1990.

Sixty fifth-grade students, 60 first-year university students, and 40 participants in a conference were exposed to four instructional different sets. Those experiencing tactually reinforced information and tactual plus visual did better on the posttest of knowledge acquired than those receiving only the visual condition. [JDS]

The Validity and Clinical Uses of the Pepper Visual Skills for Reading Test. Watson G, Baldasare J, Whittaker S, *Visual Impairm Blindn* 84:119-123, 1990.

Using the Pepper VSRT and the Gray Oral Reading Test, authors individually tested 38 persons, 21 to 91 years of age. The two reading test scores correlated 0.82, and the correlation between reading rate and Pepper score was 0.32. Authors conclude that the Pepper VSRT is, ". . . valid and reliable when used as an evaluation tool in reading for individuals with maculopathies who formerly read at a sixth-grade level." [JDS]

Visual Biasing of Normal and Impaired Auditory Speech Perception. Walden BE, Montgomery AA, Prosek RA, Hawkins DB, *J Speech Hear Res* 33:163-173, 1990.

To determine the extent to which discrepant visual cues bias auditory perception, authors presented 15 normally and 15 abnormally hearing persons with acoustic computer-generated, consonant-vowel stimuli, with and without visual cues. Persons with impaired hearing proved more susceptible to influence by the visual cues. [JDS]

BOOK REVIEWS

by
Jerome D. Schein, Ph.D.
Professor Emeritus of Sensory Rehabilitation, New York University, and David Peikoff Chair in Deafness Studies, University of Alberta, Edmonton, Canada
and
Beryl M. Benjers, Ph.D.
Departments Editor, Journal of Rehabilitation Research and Development

Clinical Decision Making in Behavior Therapy, edited by A.M. and C.M. Nezu. Champaign, IL: Research Press, 1989, 419 pp.
by Jerome D. Schein, Ph.D.

Thirty-five years after publication of Paul Meehl's classic *Clinical vs. Statistical Prediction*, it still continues to irritate clinicians who find its support of actuarial methods somehow distasteful. But the editors of the book under review recognize their debt to Meehl and, in keeping with his argument that the actuarial approach frees clinicians to spend more time in therapy, they have induced 12 contributors to join them in applying a problem-solving model to choice of treatment. The five-step model begins with Problem Orientation, progresses through Definition and Formulation, Generation of Alternatives, and Decision Making, and concludes with Solution Implementation and Verification. The cases to which the model has been applied range across agoraphobia, chronic pain, marital distress, obesity, personality disorders, post-traumatic stress, sexual dysfunction, and unipolar depression—all of which have chapters devoted to them. In addition to the decision model, child and adolescent therapy and clinical stress management are covered in separate chapters.

Cognitive Approaches in Neuropsychological Rehabilitation, edited by Xavier Seron and Gerard Deloche. Hillsdale, NJ: Lawrence Erlbaum, 1989, 411 pp. Illustrated.
by Jerome D. Schein, Ph.D.

Cognition, like all vogue terms, has become so overused and misused that its meaning has been obscured. Recognizing this pitfall, the editors note, "We call 'cognitive' those therapeutic approaches that are explicitely (*sic*) based on a model describing the organization of the cognitive apparatus viewed as a complex architecture of at least partially autonomous functional units. This therapeutic approach presumes that neuropsychological disorders reflect the total cognitive apparatus minus specific altered components. In such a perspective a therapy must be designed on the basis of hypotheses on the locus of the defective and intact processing components" (page 1). The authors of the chapters that follow appear to have adhered to this definition, insofar as it can be followed, as they address anomia, dyslexia, dysgraphia, memory, numerical transcoding, pragmatics, spontaneous recovery from aphasia, therapeutic models of cognitive therapy, unilateral spatial neglect, and use of microcomputers in therapy. The inevitable unevenness in quality of presentations inherent in a multiply-authored text is overcome, in part, by the panoramic view it affords of this field. It is largely a European product, with all but two of the authors from Belgium, England, France, and Italy. Despite this fact, the English text reads well; indeed, it enhances the value of the text for North American readers, because it introduces them to literature they do not usually peruse and because the authors reflect somewhat unfamiliar views. Another refreshing feature of the book is its final chapter, contributed by Alfonso Caramazza, who introduces himself as having ". . . no particular competence in the areas of clinical neuropsychology or therapeutic intervention," but who argues that, "The promise of cognitive neuropsychology as a guide for the choice of intervention strategies is still largely unfulfilled" (page 396). The editors merit congratulations for the inclusion of his contrary views.

Ethical Issues in Disability and Rehabilitation: Report of a 1989 International Conference, edited by Barbara Duncan and Diane E. Woods. New York: Printing Production Services, 1989, 170 pp.
by Beryl M. Benjers, Ph.D.

This volume is published as a collaborative project of the World Institute on Disability, Rehabilitation International, and the World Rehabilitation Fund, with funding from the National Institute on Disability and Rehabilitation Research. The contents are a result of the International Symposium on Ethics and Disability, held in conjunction with the Society for Disability Studies' Annual Convention on June 23–24, 1989, in Denver, Colorado, USA.

The monograph consists of five parts: 1) introductory material; 2) papers presented at this international symposium on this topic; 3) responses to the symposium, prepared by four of the participants; 4) selected additional papers which offer views from perspectives or cultures not represented at the conference; and, 5) an annotated international bibliography.

The conference papers deal with a wide variety of subjects, including genetic engineering, eugenics, decision-making projects, allocation of resources and distribution of justice, bioethics, views from the other countries, life and death decisions, decisions with respect to extremely premature and very low birthweight infants, and AIDS. In addition, selected papers on controversial issues such as high tech medical rescue intervention, issues of life and personhood, the quality of life, Nazi scientists and ethics, and euthanasia are presented.

The book also includes an interesting presentation of complicated ethical questions faced by our society today and the role of clinicians, family, clergy, courts, communities, and the severity of disability. It should be a valuable reference material for all rehabilitation and disability resource centers.

Human Communication and Its Disorders: A Review 1988, edited by Harris Winitz. Norwood, NJ: Ablex Publishing, 1988, 500 pp.
by Jerome D. Schein, Ph.D.

The second volume in this most-welcome series contains reviews of cleft palate and velopharyngeal function (D.P. Kuehn and R.M. Dalston), dysarthria (A.H.B. Putnam), and production and perception of foreign-language speech sounds (J.E. Flege). Its predecessor (Volume 1, 1987) reviewed speech perception, acquired apraxia of speech, and treatment of language disorders in children. The thoroughness of the reviews can be noted by citing two figures, the amount of text allocated to each, and the number of references cited: Kuehn and Dalston's chapter covers 106 pages and 445 citations, Putnam's has 116 pages and 193 references, and Flege's has 177 pages and 286 references. In addition, brief annotations of books on human communication and its disorders round out the second volume, providing additional citations and sources of publishers working in the area. Missing from the two volumes are introductions by the editor that give some idea of the planning underlying this series: frequency of publication, choices of topics and authors, and other related points of interest to readers.

Understanding Blindness. Mark Hollins, Hillsdale, NJ: Lawrence Erlbaum, 1989, 194 pp. Illustrated.
by Jerome D. Schein, Ph.D.

Readers seeking an introduction to the literature on blindness should be satisfied with the broad sketch of the field that Hollins provides: the nature of blindness, anatomy and physiology of the visual system, psychology of blindness (attitudes, emotions, perception and cognition) and rehabilitation. None of these topics, however, is treated in depth, leaving the sophisticated reader unhappy about matters skimmed over or ignored. For example, the author uncritically accepts the estimates of the prevalence of blindness synthesized by the National Society to Prevent Blindness, even referring to such made-up numbers as ". . . useful in that they help to guide long-term planning" (page 6). Why faulty data should be more helpful when the anticipated events are temporally distant than near is not explained. Among topics ignored or only touched upon are monocular blindness, a condition whose handicapping consequences are too often underestimated by most rehabilitators, and conjoint deafness and blindness, which is treated in two pages of text. Nonetheless, taken on its own terms as an introduction to the field, this text deserves readers. It contains a great deal of compacted information, written in an easy-to-read style and attractively presented, with clear, well-chosen illustrations.

PUBLICATIONS OF INTEREST

Compiled by Beryl M. Benjers, Ph.D.
Departments Editor

This list of references offers *Journal* readers significant information on the availability of recent rehabilitation literature in various scientific, engineering, and clinical fields. The *Journal* provides this service in an effort to fill the need for a comprehensive and interdisciplinary indexing source for rehabilitation literature.

All entries are numbered so that multidisciplinary publications may be cross-referenced. They are indicated as *See also* at the end of categories where applicable. A listing of the periodicals reported on follows the references. In addition to the periodicals covered regularly, other publications will be included when determined to be of special interest to the rehabilitation community. To obtain reprints of a particular article or report, direct your request to the appropriate contact source listed in each citation.

Page **List of Categories**

BIOENGINEERING and BIOMECHANICS

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Contact: Richard Mettler, Nebraska Dept. of Public Institutions, Division of Rehabilitation Services for the Visually Impaired, 4600 Valley Rd., Lincoln, NE 68510

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Contact: Richard G. Long, PhD, Vision Section, Sensory and Behavioral Sciences Branch, Rehabilitation Research and Development Center, VA Medical Center (Atlanta), 1670 Clairmont Rd., Decatur, GA 30033

255. Optometric Vision Therapy for Low Vision Patients. Martin SF, *J Vision Rehabil* 4(1):53-58, 1990.
Contact: Stanley F. Martin, OD, *Journal of Vision Rehabilitation*, 2440 'O' St., Suite 202, Lincoln, NE 68510

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Contact: Walter E. Needham, PhD, VA Medical Center, West Spring St., West Haven, CT 06516

257. Rehabilitative Management of Maculopathies. Kastner BD, *J Vision Rehabil* 4(1):33-47, 1990.
Contact: Bruce D. Kastner, OD, William Feinbloom Vision Rehabilitation Center, The Eye Institute, 1201 West Spencer St., Philadelphia, PA 19141

258. Self-Report in Functional Assessment of Low Vision. Szlyk JP, *et al.*, *J Visual Impairm Blindn* 84(2):61-66, 1990.
Contact: Janet P. Szlyk, PhD, The Lighthouse, 111 East 59th St., New York, NY 10022

259. Tactile Speech Indicator: Adaptive Telephone Device for Deaf-Blind Clients. Lynch MR, *J Visual Impairm Blindn* 84(1):21-22, 1990.
Contact: M.R. Lynch, 919 Walnut St., Philadelphia, PA 19107

260. Tactual Environmental Interpretation: A Multisensory Approach. Nichols DR, *J Visual Impairm Blindn* 84(3):124-125, 1990.
Contact: Douglas R. Nichols, PhD, Leisure Studies Program, University of Victoria, PO Box 1700, Victoria, British Columbia V8W 2Y2, Canada

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Contact: Gale R. Watson, MEd, Pennsylvania College of Optometry, 1200 West Godfrey Ave., Philadelphia, PA 19141

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Contact: Carma A. Heitzmann, PhD, c/o *Journal of Vision Rehabilitation*, 2440 'O' St., Suite 202, Lincoln, NE 68510

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Contact: August Colenbrander, MD, Low Vision Service, Dept. of Ophthalmology, Pacific Presbyterian Medical Center, 2340 Clay St., Room 636, San Francisco, CA 94115

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Contact: Ralph P. Garzia, OD, School of Optometry, University of Missouri-St. Louis, 8001 Natural Bridge Rd., St. Louis, MO 63121

See also 55, 74, 78, 79, 297, 301, 303

SPINAL CORD INJURY

265. Atelectasis and Pneumonia in Acute Spinal Cord Injury. Fishburn MJ, Marino RJ, Ditunno JF Jr, *Arch Phys Med Rehabil* 71(3):197-200, 1990.
Contact: Mary Jo Fishburn, MD, Good Samaritan Hospital, Professional Office Bldg., Suite 303, 5601 Loch Raven Blvd., Baltimore, MD 21239

266. Cervical Spine Injuries: A Follow-Up of 332 Patients. Ersmark H, Dalen N, Kalen R, *Paraplegia* 28(1):25-40, 1990.
Contact: H. Ersmark, MD, Dept. of Orthopaedic Surgery, Danderyd Hospital, S-182 88 Danderyd, Sweden

267. Cognitive Dysfunction in Spinal Cord Injury Patients: Sensitivity of the Functional Independence Measure Subscales versus Neuropsychologic Assessment. Davidoff GN, *et al.*, *Arch Phys Med Rehabil* 71(5):326-329, 1990.
Contact: Gary N. Davidoff, MD, Rehabilitation Medicine Service (117), VA Medical Center, 2215 Fuller Rd., Ann Arbor, MI 48105

268. Enhancement of Sexual Function and Fertility in Spinal Cord-Injured Males. Yarkony GM, *Am J Phys Med Rehabil* 69(2):81-87, 1990.
Contact: Gary M. Yarkony, MD, Rehabilitation Institute of Chicago, 345 East Superior St., Chicago, IL 60611

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Contact: Ole Slot, MD, Harald Selmersvej 50, DK-8240 Risskov, Denmark

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Contact: M. Masini, MD, Hospital for Locomotor System Disease, SARAH, Brasilia 70.330, Brazil

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Contact: P. Dollfus, MD, Centre de Readaptation, 57 rue Albert Camus, 68093 Mulhouse Cedex, France

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Contact: F. Lonnberg Christensen, Centre for Spinal Cord Injured, Fysiurgisk Hospital, Havnevej 25, DK-3100 Hornbaek, Denmark

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Contact: Svend S. Pedersen, MD, Dept. of Clinical Microbiology 8223, Rigshospitalet, Juliane Mariesvej 28.2. DK-2100 Copenhagen O, Denmark

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Contact: Finn Nykvist, MD, Rehabilitation Research Centre, Peltolantie 3, SF-20720 Turku, Finland

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Contact: J.W. Peters, MD, Dept. of Orthopaedic Surgery, Health Sciences Centre and University of Manitoba, Rm. GF311, 820 Sherbrook St., Winnipeg, Manitoba R3A 1R9, Canada

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Contact: Jay V. Subbarao, MD, Comprehensive Rehabilitation Center (11R), Edward Hines Jr. Hospital, Hines, IL 60141

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Contact: Michael B. Bracken, PhD, Dept. of Epidemiology and Public Health, Yale University School of Medicine, 60 College St., New Haven, CT 06510

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Contact: Puliyodil A. Philip, MD, Rehabilitation Institute of Chicago, 345 East Superior, Chicago, IL 60611

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Contact: Gary M. Yarkony, MD, Director of Spinal Cord Rehabilitation, Rehabilitation Institute of Chicago, 345 East Superior, Chicago, IL 60611

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Contact: Gary Davidoff, MD, Rehabilitation Medicine Service (117), VA Medical Center, 2215 Fuller Rd., Ann Arbor, MI 48105

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Contact: Semion Rochkind, MD, Dept. of Neurosurgery, Ichilov Hospital, Tel-Aviv Medical Center, Tel-Aviv, Israel

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Contact: Carleton Pilsecker, Spinal Cord Injury Service, VA Medical Center, 5901 E. 7th St., Long Beach, CA 90822

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Contact: Bianca Rucker, RN, Sexual Medicine Unit, Shaughnessy Hospital, 4500 Oak St., Vancouver, British Columbia V6H 3N1, Canada

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Contact: Thomas B. Ducker, MD, 600 Ridgely Ave., Annapolis, MD 21401

See also 33, 35, 155, 161, 163, 164, 169, 171, 177, 180, 192, 307, 313

SURGERY

285. The Acutely Injured Patient. Baker MS, *Milit Med* 155(5):215-217, 1990.
Contact: CDR Michael S. Baker, Chief of Surgery, Contra Costa County/Merrithew Memorial Hospital, 2500 Alhambra Ave., Martinez, CA 94553

286. After Triage: Lessons Learned in Vietnam with Military Casualties. Schultz RJ, *Milit Med* 155(5):221-222, 1990.
Contact: LTC R.J. Schultz, 110 East Routt Ave., Pueblo, CO 81009

287. Basic and Advanced Combat Casualty Care: A Military Problem. Heydorn WH, *Milit Med* 155(5):229-231, 1990.
Contact: COL William H. Heydorn, 40 Geldert Dr., Tiburon, CA 94920

288. Experience as the Chief of Surgery at the 67th Evacuation Hospital, Republic of Vietnam 1968 to 1969. Sebesta D, *Milit Med* 155(5):227, 1990.
Contact: LTC Donald Sebesta, PO Box 828, Davenport, WA 99122

289. A General Surgeon in Vietnam: Lessons Learned the Hard Way. Griffiths H, *Milit Med* 155(5):228-229, 1990.
Contact: COL Harold Griffiths, c/o *Military Medicine*, Association of Military Surgeons, 9320 Old Georgetown Rd., Bethesda, MD 20814

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Contact: CAPT Paul Pitlyk, 1750 El Camino Real, Burlingame, CA 94010

291. Recognition of Vascular Injury in the Trauma Patient. Baker MS, *Milit Med* 155(5):225-226, 1990.
Contact: CDR Michael S. Baker, Chief of Surgery, Contra Costa County/Merrithew Memorial Hospital, 2500 Alhambra Ave., Martinez, CA 94553

292. Training Medics for the Combat Environment of Tomorrow. Ekblad GS, *Milit Med* 155(5):232-234, 1990.
Contact: CAPT Glenn S. Ekblad, Dept. of Anesthesia, University of Michigan Hospital, Ann Arbor, MI 48109

293. Wound Ballistics: The Management of Assault Rifle Injuries. Fackler ML, *Milit Med* 155(5):222-225, 1990.
Contact: COL Martin L. Fackler, Dept. of Surgery, Letterman Army Medical Center, San Francisco, CA 94129

See also 16, 23, 137

VOCATIONAL

294. Analysis of Co-Worker Involvement in Relation to Level of Disability Versus Placement Approach Among Supported Employees. Rusch FR, Johnson JR, Hughes C, *J Assoc Persons Severe Handicaps* 15(1):32-39, 1990.
Contact: Frank R. Rusch, Transition Institute at Illinois, College of Education, 1310 South Sixth St., University of Illinois, Champaign, IL 61820

295. Employer Attitudes Toward Hiring Individuals with Mental Disabilities. Wilgosh L, Mueller HH, *Can J Rehabil* 3(1):43-47, 1989.
Contact: Lorraine Wilgosh, PhD, Dept. of Educational Psychology, Faculty of Education, 6-102 Education North, University of Alberta, Edmonton, Alberta T6G 2G5, Canada

296. The Employment Rights of People with Disabilities: A Union Perspective. Walker C, *Rehabil Dig* 20(4):14-19, 1990.
Contact: Cathy Walker, Canadian Association of Industrial Mechanical and Allied Workers, 707-12th St., New Westminster, British Columbia V3M 4J7, Canada

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Contact: Shlomo Deshen, PhD, Dept. of Sociology and Anthropology, Tel-Aviv University, Israel

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Contact: Simo Mannila, Rehabilitation Foundation, PL 39, SF-00411 Helsinki, Finland

299. The Institutionalization of Disability Myths: Impact on Vocational Rehabilitation Services. Holmes GE, Karst RH, *J Rehabil* 56(1):20-27, 1990.
Contact: Gary E. Holmes, 3636 Brush Creek Dr., Lawrence, KS 66047

300. Integrated Employment for People with Severe Physical Disabilities: Case Studies and Support Issues. Cooper A, Mank D, *Am Rehabil* 15(3):16-23, 1989.
Contact: Abby Cooper, Elliot Bay Employment Services, Seattle, WA

301. Job Roles of Rehabilitation Teachers of Blind Persons. Leja JA, *J Visual Impairm Blindn* 84(4):155-159, 1990.
Contact: James K.A. Leja, Dept. of Blind Rehabilitation, Western Michigan University, Kalamazoo, MI 49008

302. Learning Styles and Effective Teaching Technologies for Enhancing the Employment of Deaf-Blind Youth. Maxson BJ, *Int J Rehabil Res* 12(3):348-349, 1989.
Contact: B.J. Maxson, Christian Blind Mission International, Low Vision/Education Services, 218 Louisville Rd., Starkville, MS 39159

303. Paid Summer Work Experience for Rural Blind Youth. Whitman D, *J Visual Impairm Blindn* 84(2):77-78, 1990.
Contact: David Whitman, Bureau of Services for the Visually Impaired, 4300 Old Scioto Trail, Portsmouth, OH 45662

304. Pre-Employment Screening and Health Management for Safety Forces—Methods and Techniques. Mostardi RA, *et al.*, *J Orthop Sports Phys Ther* 11(9):398-401, 1990.
Contact: Richard A. Mostardi, PhD, Prof. of Biology, University of Akron, Akron, OH 44304

305. Private Sector Job Fair for People with Disabilities. Dietl D, *Worklife* 3(1):15-19, 1990.

Contact: Dick Dietl, c/o *Worklife*, Suite 636, 1111 20th St. NW, Washington, DC 20036

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Contact: W. Feldes, Industriegewerkschaft Metall, 02 Abt. Sozialpolitik, Ref. Behinderte, Postfach 11 10 31, 6000 Frankfurt/M. 11, Fed. Rep. of Germany

307. Return to Work After Spinal Cord Injury. Crisp R, *J Rehabil* 56(1):28-35, 1990.
Contact: Ross Crisp, 67 Bayswater Rd., Croydon, Victoria 3136, Australia

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Contact: Randie M. Black-Schaffer, MD, Dept. of Physiatry, New England Rehabilitation Hospital, 2 Rehabilitation Way, Woburn, MA 01801

309. Supported and Time-Limited Transitional Employment Services. Hanley-Maxwell C, *et al.*, *J Visual Impairm Blindn* 84(4):160-166, 1990.
Contact: Cheryl Hanley-Maxwell, PhD, Rehabilitation Institute, Southern Illinois University, Carbondale, IL 62901-4609

310. Vocational Evaluation in Supported Employment. Rogan P, Hagner D, *J Rehabil* 56(1):45-51, 1990.
Contact: Pat Rogan, Syracuse University, Division of Special Education and Rehabilitation, 805 Crouse Ave., Syracuse, NY 13244-2280

311. The World of Work: Is There a Match between Vocational Preparation and Employer Expectations? Wilgosh L, *et al.*, *Can J Rehabil* 3(2):113-118, 1989.
Contact: Lorraine Wilgosh, PhD, Dept. of Educational Psychology, 6-102 Education North, University of Alberta, Edmonton, Alberta T6G 2G5, Canada

See also 251, 256

WHEELCHAIRS and POWERED VEHICLES

312. 8th Annual Survey of the Lightweights. Sunderlin A, *Sports 'N Spokes* 15(6):25-52, 1990.
Contact: Ann Sunderlin, c/o Sports 'N Spokes, 5201 North 19th Ave., Suite 111, Phoenix, AZ 85015

313. The Choice of Self-Propelling Wheelchairs for Spinal Patients. Fyfe NCM, Wood J, *Clin Rehabil* 4(1):51-56, 1990.
Contact: Dr. N.C.M. Fyfe, Disablement Services Centre, Freeman Rd., Newcastle upon Tyne NE7 7AF, UK

314. Development of a Realistic Method to Assess Wheelchair Propulsion by Disabled People. Mattison PG, Hunter J, Spence S, *Int J Rehabil Res* 12(2):137-145, 1989.
Contact: Dr. Paul G. Mattison, Rehabilitation Studies Unit, Princess Margaret Rose Hospital, Fairmilehead, Edinburgh EH10 7ED, Scotland

315. The Wheelchair Simulator: Powered Wheelchair Assessment and Training. Scott-Taplin C, *et al.*, *Can J Rehabil* 3(1):49-51, 1989.
Contact: Cathie Scott-Taplin, The Technical Resource Centre, #200, 1201-5th St. SW, Calgary T2R 0Y6, Canada

See also 173, 213, 242

Periodicals reviewed for PUBLICATIONS OF INTEREST

Accent on Living
Acta Orthopaedica Scandinavica
Advances in Orthopaedic Surgery
American Annals of the Deaf
American Journal of Occupational Therapy
American Journal of Ophthalmology
American Journal of Optometry and Physiological Optics
American Journal of Physical Medicine and Rehabilitation
American Journal of Sports Medicine
American Rehabilitation
Annals of Biomedical Engineering
AOPA Almanac (American Orthotic and Prosthetic Association)
Applied Optics
Archives of Physical Medicine and Rehabilitation
ASHA (American Speech and Hearing Association)
Biomaterials, Artificial Cells and Artificial Organs
Biomedical Instrumentation and Technology
Biomedical Technology Today
Biotechnology and Bioengineering

British Journal of Occupational Therapy
Caliper (Canadian Paraplegic Association)
Canadian Journal of Rehabilitation
Clinical Biomechanics
Clinical Kinesiology
Clinical Orthopaedics and Related Research
Clinical Physics and Physiological Measurement
Clinical Prosthetics and Orthotics
Clinical Rehabilitation
Communication Outlook
CRC Critical Reviews in Biomedical Engineering
DAV Magazine (Disabled American Veterans)
Discover
Electromyography and Clinical Neurophysiology
Electronic Design
Electronic Engineering
Electronics
Engineering in Medicine
Ergonomics
Harvard Medical School Newsletter
Hearing Journal
Hearing Research
Human Factors: The Journal of the Human Factors Society
IEEE Engineering in Medicine and Biology Magazine
IEEE Transactions in Biomedical Engineering
IEEE Transactions in Systems, Man and Cybernetics
International Disability Studies
International Journal of Rehabilitation Research
International Journal of Technology & Aging
International Rehabilitation
Journal of Acoustical Society of America
Journal of American Optometric Association
Journal of Biomechanical Engineering
Journal of Biomechanics
Journal of Biomedical Engineering
Journal of Biomedical Materials Research
Journal of Bone and Joint Surgery—American Ed.
Journal of Bone and Joint Surgery—British Ed.
Journal of Clinical Engineering
Journal of Head Trauma and Rehabilitation
Journal of Medical Engineering and Technology
Journal of Neurologic Rehabilitation
Journal of Optical Society of America A
Journal of Orthopaedic and Sports Physical Therapy
Journal of Orthopaedic Research
Journal of Physics D: Applied Physics
Journal of Rehabilitation
Journal of Speech and Hearing Disorders
Journal of Vision Rehabilitation
Journal of Visual Impairment and Blindness
Laser Focus
Mayo Clinic Proceedings
Medical and Biological Engineering and Computing
Medical Device and Diagnostic Industry
Medical Electronics
Medical Instrumentation
Medical Physics
Medical Progress Through Technology
Medical Psychotherapy
Medicine & Science in Sports and Exercise
Military Medicine
New England Journal of Medicine
The Occupational Therapy Journal of Research
Orthopaedic Review
Orthopedic Clinics of North America
Orthopedics
Orthotics and Prosthetics
Palaestra
Paraplegia
Paraplegia News
Physical and Occupational Therapy in Geriatrics
Physical Therapy
Physical Medicine and Rehabilitation
Physics Today
Physiotherapy
Proceedings of the Institution of Mechanical Engineers—Part H: Journal of Engineering in Medicine
Prosthetics and Orthotics International
Rehabilitation Digest
Rehabilitation Gazette
Robotics World
Scandinavian Journal of Rehabilitation Medicine
Science
Science News
Scientific American
Speech Technology
Spine
SOMA: Engineering for the Human Body
Sports 'N Spokes
Technical Aid to the Disabled Journal
Techniques in Orthopaedics
Together
Topics in Geriatric Rehabilitation
VA Practitioner
Volta Review
Worklife

CALENDAR OF EVENTS

Compiled by Beryl M. Benjers, Ph.D.
Departments Editor

NOTE: An asterisk at the end of a citation indicates a new entry to the calendar.

1990

August 5-9, 1990
American Association of Physicists in Medicine (AAPM), 32nd Annual Meeting, St. Louis, MO
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

August 5-10, 1990
10th World Congress of the International League of Societies for Persons with Mental Handicap (ILSMH), Paris, France
Contact: Secretariat, ILSMH, 248, Avenue Louise, Bte. 17, B-1050 Brussels, Belgium

August 6-10, 1990
Gordon Research Conference on Bioengineering and Orthopaedic Science, New Hampshire
Contact: Alan J. Grodzinsky, Room 38-377, MIT, Cambridge, MA 02139*

August 12-15, 1990
Fourth Annual Summer Speech/Hearing/Language Seminars, South Lake Tahoe, NV
Contact: Robert R. Shearer, PO Box 567, Concord, CA 94522-0567*

August 12-16, 1990
8th International Congress of ISEK, Baltimore, MD
Contact: ISEK 1990, Congress Secretariat, PO Box 986, Hunt Valley, MD 21030*

August 12-16, 1990
Fourth International Conference of the International Society for Augmentative and Alternative Communication (ISAAC), Stockholm, Sweden
Contact: ISAAC 1990, c/o Destination Stockholm, Stora Hoparegrand 5, S-111 30 Stockholm, Sweden or The Institute for Integration, Norrmalmstorg 1, 114 46 Stockholm, Sweden; Tel 468 112006

August 20-23, 1990
9th International Conference Medical Informatics Europe (MIE) '90 (EFMI/BCS), Glasgow, UK
Contact: MIE Secretariat, Meeting Makers, 50 Richmond St., Glasgow G1 1XP, UK*

August 21-23, 1990
3rd International Conference Biointeractions '90, St. Catherine's College, Oxford, UK
Contact: Helen Juden, Conference Organizer, Biointeractions '90, Butterworth Scientific Ltd., PO Box 63, Westbury House, Bury St., Guildford, Surrey GU2 5BH, UK*

August 23-26, 1990
International Conference on Comprehensive System of Social and Other Services for Disabled Persons, Warsaw, Poland
Contact: Polish Society for Rehabilitation of the Disabled, ul. Partyzantow 4 m.10, 00629 Warszawa, Poland*

August 26-28, 1990
East Coast Conference on Biomechanics, Old Westbury, NY
Contact: Prof. H.S. Ranu, Dept. of Biomechanics, NYCOM, New York Institute of Technology, Old Westbury, NY 11568; (516) 626-6926, Fax (516) 626-1306

August 26-29, 1990
World Congress on Health Technology Standards, Dublin, Ireland
Contact: Electro-Technical Council of Ireland, 1 Fitzwilliam Pl, Dublin 2, Ireland*

August 27-30, 1990
International Symposium on Signal Processing and Its Application, Queensland, Australia
Contact: Uniquest, Ltd., University of Queensland, St. Lucia QLD 4067, Australia

August 27-September 1, 1990
10th International Symposium on External Control of Human Extremities, Dubrovnik, Yugoslavia
Contact: D. Popovic, ETAN, PO Box 356, 11001 Belgrade, Yugoslavia

August 29-31, 1990
6th IMEKO Conference on Measurement in Clinical Medicine and 8th Hungarian Conference on Biomedical Engineering, Sopron, Hungary
Contact: Dr. Gyorgy Bogany, Merestechnikai es Automatizalasi Tudomanyos Egyesulet, POB 457, H-1372 Budapest, Hungary

August 30-September 4, 1990
First World Congress of Biomechanics, University of California-San Diego, La Jolla, CA
Contact: Lene Hartman, Conf. Mgr., AMES-Bioeng. R-012, UC-San Diego, La Jolla, CA 92093; (619) 534-4272

September 3-7, 1990
Fracture Bracing, Glasgow, Scotland
Contact: Prof. J. Hughes, National Centre for Training and Education in Prosthetics and Orthotics, University of Strathclyde, Curran Bldg., 131 St. James' Rd., Glasgow G4 0LS, Scotland

September 4-6, 1990
IVth International Aphasia Rehabilitation Congress, Edinburgh, Scotland
Contact: Aphasia Rehabilitation Congress, Capital Campus, Queen Margaret College, Edinburgh EH12 8TS, Scotland

September 4-7, 1990
International Centre for Theoretical Physics (ICTP), Istituto Nazionale di Fisica Nucleare (INFN), and Societa Italiana di Fisica (SIF), 3rd International Conference on "Applications of Physics in Medicine and Biology, Medical Diagnostic Imaging," Miramare, Trieste, Italy
Contact: ICTP, Conference on Application of Physics in Medicine and Biology, PO Box 586, I-34100 Trieste, Italy*

September 5-8, 1990
European Congress of Laryngectomees, Amsterdam, The Netherlands
Contact: Congress Secretariat, RAI Organisatie Bureau Amsterdam bv, Europaplein 12, 1078 GZ Amsterdam, The Netherlands; Tel (31) 20-5491212, Telefax (31) 20-464469, Telex 13499*

September 6-7, 8, 1990
International Society for Fracture Repair (ISFR-90), Rochester, Minneapolis, MN
Contact: Conference Secretariat ISFR-90, Mrs. Joanne Murray, Mayo Clinic/Mayo Foundation, Orthopedics Biomechanics Lab., Guggenheim Bldg., C-053, Rochester, MN 55905; (507) 284-2262, Fax (507) 284-5392

September 9-11, 1990
New York State Association of Rehabilitation Facilities (NYSARF), 12th Annual Conference and Trade Show, "Redefining Rehab: Initiating a Decade of Change," Albany, New York
Contact: NYSARF, 155 Washington Ave., Suite 305, Albany, NY 12210; (518) 449-2976*

September 9-14, 1990
Focus on Independence: Trinity Seminars, Geriatrics 1990, Dublin, Ireland
Contact: Trinity Seminars, Betty Cox Associates, Inc., 232 East University Pkwy., Baltimore, MD 21218; (301) 243-5943*

September 9-14, 1990
Interfaces 90: International Conference on Interfaces in Medicine and Mechanics, Bologna, Italy
Contact: Dr. K.R. Williams, Dept. of Basic Dental Science, University of Wales, College of Medicine, Heath Park, Cardiff, Wales CF4 4XW, UK

September 9-15, 1990
International Skeletal Society 17th Annual Refresher Course, Salzburg, Austria
Contact: Secretary/Treasurer, International Skeletal Society, Radiology, Hospital of the University of Pennsylvania, 3400 Spruce St., Philadelphia, PA 19104

September 10-13, 1990
MICRO SYSTEMS 1990, Berlin, Germany
Contact: MESAGO, Messe & Kongreb GmbH, Postfach 103261, D-7000 Stuttgart 10, FRG*

September 10-14, 1990
13th International Symposium HFT '90, Human Factors in Telecommunications, Torino, Italy
Contact: Roberto Marion, CSELT, PO Box 145, Torino Centro, 10100 Torino, Italy; Tel +39 11 2169755, Fax +39 11 2169520, Telex 220539 cselt

September 11-12, 1990
Biomechanical Stress Analysis (IOP/BES), Sheffield, UK
Contact: Mr. C. Jones, Meetings Officer, The Institute of Physics, 47 Belgrave Square, London SW1X 8QX, UK*

September 11-16, 1990
American Orthotic and Prosthetic Association (AOPA) Annual National Assembly, Boston, MA
Contact: Katie Register, AOPA National Headquarters, 717 Pendleton St., Alexandria, VA 22314; (703) 836-7116

Sept. 12-14, 1990
British Orthopaedic Association (BOA), Birmingham, UK
Contact: BOA, 35-43 Lincoln's Inn Fields, London WC2A 3PN, UK*

September 12-15, 1990
REHAB '90: 6th International Exhibition and Rehabilitation Aid for Further Education, Karlsruhe, West Germany
Contact: Christiana Hennemann, PO Box 100 555 D-4600, Dortmund, West Germany

September 12-16, 1990
Institute of Physical Sciences in Medicine Annual Conference, Oxford, England
Contact: Institute of Physical Sciences in Medicine, 2 Low Ousegate, York YO1 1QU, UK*

September 13-16, 1990
Cognitive Rehabilitation and Community Integration: A Focus on Educational, Family and Vocational Issues After Brain Injury, Richmond, VA
Contact: Jeffrey S. Kreutzer, PhD, Director, Rehabilitation Psychology and Neuropsychology, Medical College of Virginia, Box 677, Richmond, VA 23298-0677; (804) 786-0200

September 17-19, 1990
30th AGM Biological Engineering Society (BES), Durham, UK
Contact: Mrs. B. Freeman, BES, RCS, 35/43 Lincoln's Inn Fields, London WC2A 3PN, UK

September 17-22, 1990
VIIth International Congress on Neuromuscular Diseases (German Association Combatting Muscle Diseases), Munich, West Germany
Contact: Frank Lehmann-Horn, MD, Dept. of Neurology and Clinical Neurophysiology, Technical University of Munich, Mohlstrasse 28, D-8000 Munich 80, West Germany

September 18-21, 1990
European Signal Processing Conference, Barcelona, Spain
Contact: Eusipco—90 Scientific Committee, Dept. Teona de la Senal y Communicaciones, ETSITB-UPC, Apdo 30002, 08080 Barcelona, Spain

September 19-21, 1990
The First National Conference on Continuing Education for Adults with a Visual Impairment, entitled "Choice and Challenge," Bristol Polytechnic, England
Contact: Dr. D.J. Hill, Dept. of Continuing Education, Wills Memorial Building, Queens Rd., Bristol BS8 1HR England*

September 19-21, 1990
30th AGM of the Biological Engineering Society, Durham, England
Contact: Mrs. B. Freeman, BES, 35 Lincoln's Inn Fields, London, England*

September 19-22, 1990
XVII Congress of the European Society for Artificial Organs (ESAO), Bologna, Italy
Contact: Prof. V. Bonomini, President XVII ESAO Congress, Institute of Nephrology, St. Orsola University Hospital, Via Massarenti 9, 40138 Bologna, Italy

September 20-22, 1990
International Society for Prosthetics and Orthotics (ISPO), Australian National Member Society (ANMS), Annual Scientific Meeting, Greenslopes, Queensland, Australia
Contact: Valma Angliss, Hon. Sec., ISPO, Central Development Unit, Heidelberg Repatriation General Hospital, West Heidelberg 3081 Vic., Australia; (03) 499 6099 or Ms. Belle Davis, Physiotherapy Dept., Royal Brisbane Hospital, (07) 253 8412 or Mr. Rod Goodrick, RALAC, Greenslopes, Queensland, Australia; (07) 394 7262

September 20-30, 1990
Festival for Hearing, Erlangen, West Germany
Contact: Dr. Wolfgang Sowa, Palais Stutterheim, Marktplatz, D-8520, Erlangen, West Germany; Tel 9131/86-2735*

September 23-26, 1990
International Conference on "Comprehensive System of Social and Other Services for Disabled Persons," Warsaw, Poland
Contact: Polish Society for Rehabilitation of the Disabled, ul. Partyzantow 4 m.10, 00-629 Warszawa, Poland

September 23-26, 1990
2nd International Congress on Tissue Integration in Oral, Orthopedic and Maxillofacial Reconstruction, Mayo Medical Center, Rochester, MN
Contact: 2nd International Tissue Integration Congress, Dept. of Dentistry, Mayo Clinic, Rochester, MN 55905

September 27-29, 1990
Chartered Society of Physiotherapy, Annual Congress 1990, "Forward to Basics," Bournemouth, England
Contact: Chairman, 1990 Programme Organising Committee, c/o Events Organiser, Chartered Society of Physiotherapy, 14 Bedford Row, London WC1R 4ED, England

September 30-October 1-3, 1990
4th International Francophone Congress on Gerontology, Montreal, Quebec, Canada
Contact: Ms. Carolyn Sharp, Director of Communications, Secretariat du congres: Les Services de congres GEMS, 4260 Girouard, Bureau 100, Montreal, Quebec H4A 3C9, Canada; (514) 485-0855, Fax (514) 487-6725

October 5-7, 1990
Take Control: The Adaptive Technology Conference (Courses, Workshops and Hands-On Training in Computer Applications for the Disabled), St. Louis, MO
Contact: Take Control, Madalaine Pugliese Associates, Inc., 5 Bessom St., Suite 175, Marblehead, MA 01945; (617) 639-1930

October 8-12, 1990
Human Factors Society, 34th Annual Meeting, Orlando, FL
Contact: Human Factors Society, PO Box 1369, Santa Monica, CA 90406; (213) 394-1811/9793

October 10-11, 1990
Rehabex '90 Show and Conference, Toronto, Canada
Contact: ECM, 1599 Hunrontario St., Suite 301, Missauga, Ontario L5G 3H7, Canada; (416) 274-5505*

October 11-13, 1990
First IFMBE Far Eastern Conference on Medical and Biological Engineering, Tokyo, Japan
Contact: Prof. Chiyoshi Yoshimoto, West 12, South 17, Sapporo 064, Hokkaido, Japan

October 12-13, 1990
New York Chapter of the American Academy of Orthotists and Prosthetists (AAOP) Fall Meeting, Binghamton, New York
Contact: Bryan Finley, CP, (607) 797-1246*

October 14-17, 1990
Bioelectrical Repair and Growth Society (BRAGS), 10th Anniversary Symposium and Meeting, Philadelphia, PA
Contact: Executive Secretary, BRAGS, PO Box 64, Dresher, PA 19025; (215) 659-5180*

October 14-18, 1990
International Congress of Audiology, Tenerife, Canary Islands, Spain
Contact: J.J. Barajas, Perez de Rozas, 8-38004 Santa Cruz de Tenerife, Canary Islands, Spain; Tel (22) 27 54 88, Fax (22) 27 03 64*

October 17-21, 1990
Eastern Orthopedic Association, Southampton, Bermuda
Contact: Eastern Orthopedic Association, 301 South 8th St., Suite 3F, Philadelphia, PA 19106*

October 19-20, 1990
American Academy of Orthotists and Prosthetists (AAOP), Continuing Education Conference 3-90 and North Carolina Chapter, North and South Carolina Societies of Orthotists and Prosthetists Combined Meeting, "Sports and Recreational Prosthetics," Research Triangle Park, NC
Contact: AAOP National Office, (703) 836-7118

October 21-26, 1990
American Congress of Rehabilitation Medicine, American Academy of Physical Medicine and Rehabilitation, 67th Annual Session, Phoenix, AZ
Contact: American Academy of Physical Medicine and

Rehabilitation, 78 East Adams St., Chicago, IL 60603; (313) 922-9371 or 3613*

October 22-26, 1990
International Conference on Signal Processing, Beijing, China
Contact: Prof. Yuan Baozong, Research Institute of Information Science, Northern Jiaotong University, Beijing 100044, China*

October 23-25, 1990
3rd International Biotechnology Expo and Scientific Conference, San Mateo, CA
Contact: IBEX, 3097 Moorpark Ave., Suite 202, San Jose, CA 95128; (408) 554-6644*

October 26-30, 1990
9th Asia/Pacific Rehabilitation International Conference, Beijing, China
Contact: J. Morrison, RADAR, 25 Mortimer St., London W1N 8AB, UK*

October 28-31, 1990
12th National Conference on Specialized Transportation, Sarasota, FL
Contact: Mr. James Scott, Staff Coordinator, 12th National Conference on Specialized Transportation, Transportation Research Board, 2101 Constitution Ave. NW, Washington, DC 20418

October 28-November 2, 1990
Society for Neuroscience Annual Meeting, St. Louis, MO
Contact: Nancy Beamg, Society for Neuroscience, 11 Dupont Circle NW, Suite 500, Washington, DC 20036*

October 29-31, 1990
Functional Electrical Stimulation: Practical Aspects for Clinicians, Glasgow, Scotland
Contact: Dr. C.A. Kirkwood, Bioengineering Unit, Wolfson Centre, 106 Rottenrow, University of Strathclyde, Glasgow G4 0NW, Scotland*

November 1-3, 1990
The London International Symposium on Arthroscopy, London, UK
Contact: MetaPhor Conferences & Meetings, 21 Kirklees Close, Farsley, Pudsey, West Yorkshire LS28 5TF, UK; Tel 44-532-550752, Fax 44-532-571495

November 1-4, 1990
IEEE Engineering In Medicine and Biology Society (IEEE/EMBS), 12 Annual International Conference, Philadelphia, PA
Contact: IEEE, 445 Hoes Lane, Piscataway, NJ 08855; (201) 981-0060

November 1-4, 1990
Speech Communication Association, Chicago, IL
Contact: Speech Communication Association (SCA), 5105 Backlick Rd., Annandale, VA 22003*

November 4-7, 1990
Symposium on Computer Applications in Medical Care (SCAMC), Washington, DC
Contact: SCAMC, The George Washington University Medical Center, Office of Continuing Medical Education, 2300 K St. NW, Washington, DC 20037

November 4-8, 1990
International Symposium on Sexuality and Disabilities, Tel Aviv, Israel
Contact: E. Chigier, MD, Israel Rehabilitation Society, 18 David Elazar St., Tel Aviv, 61909 Israel*

November 5-8, 1990
European Conference on the Advancement of Rehabilitation Technology (ECART), Maastricht, The Netherlands
Contact: Dr. Th. Gerritsen/Dr. Ir. M. Soede, Institute for Rehabilitation Research, Zandbergsweg 111, 6432 CC Hoensbroek, The Netherlands; Tel 045-224300

November 9-11, 1990
Rehab '90—"Rehabilitation: Creating Opportunities," Minneapolis, MN
Contact: National Rehabilitation Association, 633 South Washington St., Alexandria, VA 22314-4193; (703) 836-0850/0852 TDD*

November 11-14, 1990
Xth International Conference on the Use of Computers in Radiotherapy, Lucknow, India
Contact: Dr. S. Hukku, Dept. Radiotherapy, Sanjay Gandhi P.G.I., Rai Bareli Rd., PO Box 375, Lucknow 226001, India; Tel 0522 54336 Telex 535-411

November 13-17, 1990
National Head Injury Foundation (NHIF) Ninth Annual

National Symposium, Head Injury Frontiers—1990, New Orleans, LA
Contact: Symposium Committee, NHIF, 333 Turnpike Rd., Southborough, MA 01772; (508) 485-9950

November 15-16, 1990
Fourteenth Annual Meeting of the American Society of Biomechanics, University of Miami, Coral Gables, FL
Contact: T.M. Khalil, Dept. of Industrial Engineering, University of Miami, PO Box 248294, Coral Gables, FL 33124-0623; (305) 284-2344*

November 16-19, 1990
American Speech-Language-Hearing Association (ASHA), Annual Convention, Seattle, WA
Contact: ASHA, 10801 Rockville Pike, Rockville, MD 20850; (301) 897-5700

November 17-18, 1990
Ninth Southern Biomedical Engineering Conference, Miami, FL
Contact: Dr. Gautam Ray, Conference Chairman, Mechanical Engineering Dept., The State University of Florida at Miami, University Park, Miami, FL 33199; (305) 554-2569

November 19-22, 1990
North Sea Conference Biomedical Engineering 1990, Antwerp, Belgium
Contact: Technologisch Instituut-K VIV, Attn: Mr. Luk Pauwels, Desguinlei 214, B-2018 Antwerp, Belgium; Tel 32 3 216 09 96, Fax 32 3 216 06 89

November 21-23, 1990
Cancer Care: The Complete Circle, Edmonton, Alberta, Canada
Contact: Shaunne Letourneau, Boyle, Letourneau & Assoc. Inc., 4 Lucerne Crescent, St. Albert, Alberta T8N 2R2, Canada

November 25-30, 1990
Joint Meeting of the American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America (RSNA), Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

November 26-30, 1990
Acoustical Society of America (ASA), San Diego, CA
Contact: ASA, 335 East 45th St., New York, NY 10017

November 28, 1990
Traumatic Brain Injury in the Pediatric Population: Community/Academic Reintegration, Mountainside, NJ
Contact: Sallie Comey, Children's Specialized Hospital, New Providence Rd., Mountainside, NJ 07091; (201) 233-3720*

December 4-9, 1990
American Academy of Neurological and Orthopaedic Surgeons, Las Vegas, NV
Contact: M.R. Rask, MD, 2320 Rancho Dr., Suite 108, Las Vegas, NV 89102*

December 6-8, 1990
Sixth International Conference on Biomedical Engineering, Singapore
Contact: The Secretary, Sixth Biomed Conference 1990, Dept. of Orthopaedic Surgery, National University Hospital, 5 Lower Kent Ridge Rd., Singapore 0511, Republic of Singapore; Tel 7724423/7724327, Telex RS 55503 NUH, Fax 7795678

December 11-15, 1990
Fitting Procedures for the Utah Artificial Arm and Hand Control, UCLA Prosthetic Education Program, Los Angeles, CA
Contact: Motion Control Home Office, 1290 West 2320 South, Suite A, Salt Lake City, UT 84119; 1-800-621-3347

1991

February 14-19, 1991
American Association for the Advancement of Science (AAAS) Annual Meeting, Washington, DC
Contact: AAAS Meeting Office, 1333 H St. NW, Washington, DC 20005; (202) 326-6450*

March 7-12, 1991
American Academy of Orthopaedic Surgeons (AAOS) Annual Meeting, Anaheim, CA
Contact: AAOS, (312) 823-7186

March 18-20, 1991
American Spinal Injury Association, Seattle, WA
Contact: American Spinal Injury Association, 250 East Superior St., Room 619, Chicago, IL 60611*

March 19-24, 1991
American Academy of Orthotists and Prosthetists (AAOP), Annual Meeting and Scientific Symposium,

San Diego, CA
Contact: AAOP National Office, (703) 836-7118

March 20-23, 1991
The Sixth Annual International Conference, "Technology and Persons with Disabilities," Los Angeles, CA
Contact: Dr. Harry J. Murphy, Office of Disabled Student Services, California State University, Northridge, 18111 Nordhoff St., Northridge, CA 91330; (818) 885-2578, Fax (818) 885-4545*

April 8-10, 1991
American Spinal Injuries Association (ASIA), Seattle, WA
Contact: Lesley M. Hudson, MA, 2020 Peachtree Rd. NW, Atlanta, GA 30309

April 24-26, 1991
9th International Congress of the World Federation of Spine Surgeons and Spondyliatrists, The Hague, The Netherlands
Contact: Dr. Tonino, De Wever Ziekenhuis, H. Dunantstraat 15, 6419 PC Heerlen, The Netherlands

April 24-27, 1991
National Braille Association, Allentown, PA
Contact: National Braille Association, 1290 University Ave., Rochester, NY 14607*

April 24-28, 1991
Mid-America Orthopaedic Association, Palm Springs, CA
Contact: Dr. E.W. Johnson, Jr., 411 First Bank Bldg., Rochester, MN 55902*

May 16-18, 1991
American Orthotic and Prosthetic Association (AOPA), Region V Annual Meeting, Columbus, OH
Contact: Don Peters, (419) 522-0055

June 3-6, 1991
American Orthopaedic Association (AOA) Annual Meeting, Palm Beach, FL
Contact: AOA, 222 S. Prospect Ave., Park Ridge, IL 60068*

June 17-20, 1991
Canadian Organization of Medical Physicists Annual Meeting with Canadian College of Physicists in Medicine and Canadian Radiation Protection Association, Winnipeg, Manitoba, Canada
Contact: Dr. Walter Huda, Medical Physics, 100 Olivia St., Winnipeg, Manitoba R3E 0V9, Canada; (204) 787-4191 or Fax (204) 783-6875*

June 22-26, 1991
RESNA Rehabilitation Technology, 14th Annual Conference, Kansas City, MO
Contact: RESNA, Association for the Advancement of Rehabilitation Technology, Suite 700, 1101 Connecticut Ave. NW, Washington, DC 20036; (202) 857-1199

June 23-27, 1991
Annual Conference of the American Physical Therapy Association (APTA), Boston, MA
Contact: Information Dept., APTA, 1111 N. Fairfax St., Alexandria, VA 22314

July 7-12, 1991
World Congress on Medical Physics and Biomedical Engineering, Kyoto, Japan
Contact: Japan Convention Services Inc., Osaka Branch, Sumitomo Seimei Midosuji Bldg., 4-14-3, Nishitemma, Kita-ku, Osaka 530, Japan

July 21-25, 1991
American Association of Physicists in Medicine (AAPM), 33rd Annual Meeting, San Francisco, CA
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

August 18-23, 1991
The Fifth Congress of the International Psychogeriatric Association (IPA), Jerusalem, Israel
Contact: IPA Fifth Congress, PO Box 50006, Tel Aviv 61500, Israel*

September 2-6, 1991
6th Meeting World Federation for Ultrasound in Medicine and Biology, Copenhagen, Denmark
Contact: Soren Hanke, Ultralydlaboratoriet, Kobenhavns Amts Sygehus, Gentofte, DK-2900 Hellerup, Denmark*

September 6-8, 1991
2nd Scientific Meeting of the Scandinavian Medical Society of Paraplegia, Copenhagen, Denmark
Contact: Centre for Spinal Cord Injured, Rigshospitalet, TH2002, Blegdamsvej 9, DK-2100 Copenhagen, Denmark; Tel (+45)31 38 66 33, Ext. 2007

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September 16-20, 1991
DUNDEE '91—International Conference and Instructional Course on Orthotics, Dundee, Scotland
Contact: Dundee '91 Secretariat, c/o Dundee Limb Fitting Centre, 133 Queen St., Broughty Ferry, Dundee, DD5 1AG, Scotland*

October 1-6, 1991
American Orthotic and Prosthetic Association (AOPA), Annual National Assembly, Anaheim, CA
Contact: AOPA National Headquarters, 717 Pendleton St., Alexandria, VA 22314; (703) 836-7116

October 13-16, 1991
7th International Conference on Mechanics in Medicine and Biology, Ljubljana, Yugoslavia
Contact: ICMMB 91, Technical Organiser, CANKARJEV DOM, Cultural and Congress Centre, Kidricev park 1,61000 Ljubljana, Yugoslavia*

October 21-26, 1991
AOPA Annual National Assembly, Anaheim, CA
Contact: AOPA Headquarters, 717 Pendleton St., Alexandria, VA 22314, (703) 836-7116

October 28-30, 1991
New Hampshire Chapter of American Physical Therapy Association, "Biomechanical Evaluation and Treatment of Foot and Ankle Dysfunction," Concord, NH
Contact: Diane Spahn, (603) 437-1026, Tuesdays through Fridays

November 4-8, 1991
Acoustical Society of America (ASA), Houston, TX
Contact: ASA, 335 East 45th St., New York, NY 10017

November 13-15, 1991
ACCESS EXPO, Conference and Exposition of Accessibility, Washington, DC
Contact: The Fairfield Factor, Inc., Access Expo Public Relations Counsel, 13 Obtuse Rocks Rd., Brookfield, CT 06804; Art Kerley, (203) 775-0422*

November 17-20, 1991
15th Symposium on Computer Applications in Medical Care (SCAMC), Washington, DC
Contact: The George Washington University Medical Center, Office of Continuing Education, 2300 K St. NW, Washington, DC 20037

November 17-22, 1991
Joint Meeting of the American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America, Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

November 22-25, 1991
American Speech-Language-Hearing Association (ASHA), Annual Convention, Atlanta, GA
Contact: Conventions, (301) 897-5700

December 1-6, 1991
Joint Meeting of American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America (RSNA), Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

December 9-13, 1991
13th International Congress on Biomechanics, Perth, Western Australia
Contact: 13th Congress Secretariat, Dept. of Human Movement, The University of Western Australia, Nedlands WA 6009, Australia

1992

February 20-25, 1992
American Academy of Orthopaedic Surgeons (AAOS) Annual Meeting, Washington, DC
Contact: AAOS, (312) 823-7186

April 7-12, 1992
American Academy of Orthotists and Prosthetists (AAOP), Annual Meeting and Scientific Symposium, Miami, FL
Contact: AAOP National Office, (703) 836-7118

August 23-27, 1992
American Association of Physicists in Medicine (AAPM), 34th Annual Meeting with the Division of Medical and Biological Physics of the Canadian Association of Physicists, Calgary, Alberta, Canada
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

October 26-31, 1992
American Orthotic and Prosthetic Association (AOPA), Annual National Assembly, Orlando, FL
Contact: Katie Register, (703) 836-7116

November 8-11, 1992
16th Symposium on Computer Applications in Medical Care (SCAMC), Baltimore Convention Center, Baltimore, MD
Contact: The George Washington University Medical Center, Office of Continuing Education, 2300 K St. NW, Washington, DC 20037*

November 20-23, 1992
American Speech-Language-Hearing Association (ASHA) Annual Convention, San Antonio, TX
Contact: Conventions: (301) 897-5700*

November 29-December 4, 1992
Joint Meeting of the American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America (RSNA), Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

1993

February 18-23, 1993
American Academy of Orthopaedic Surgeons (AAOS) Annual Meeting, San Francisco, CA
Contact: AAOS, (312) 823-7186

March 30-April 4, 1993
American Academy of Orthotists and Prosthetists (AAOP), Annual Meeting and Scientific Symposium, Las Vegas, NV
Contact: AAOP National Office, (703) 836-7116

August 8-12, 1993
35th Annual Meeting of the American Association of Physicists in Medicine (AAPM), Washington, DC
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

October 25-30, 1993
American Orthotic and Prosthetic Association (AOPA), Annual National Assembly, Phoenix, AZ
Contact: Katie Register, (703) 836-7116

November 28-December 3, 1993
Joint Meeting of the American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America (RSNA), Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404*

1994

July 24-28, 1994
American Association of Physicists in Medicine (AAPM), 36th Annual Meeting, Anaheim, CA
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404*

November 27-December 2, 1994
Joint Meeting of the American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America (RSNA), Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404*

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